A HOME-BASED REHABILITATION SYSTEM FOR DEFICIENT KNEE PATIENTS

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Abstract

The Smart Health paradigm has opened up immense possibilities for designing cyber-physical systems with integrated sensing and analysis for data-driven healthcare decision-making. Clinical motor-rehabilitation has traditionally tended to entail labor-intensive approaches with limited quantitative methods and numerous logistics deployment challenges. We believe such labor-intensive rehabilitation procedures offer a fertile application field for robotics and automation technologies which is easily applicable to home-based rehabilitation system.

Our long-term goal is the creation, analysis and validation of a Home-based Rehabilitation Framework comprised of quantitative human subject measurement technologies, an adjustable smart brace coupled with an integrated PC-based control system to enhance rehabilitation process for deficient knee patient.

Human motion-capture and computational analysis tools have played a significant role in a variety of product-design and ergonomics settings for over a quarter-century. However, there exist significant differences in the capabilities and ease-of-use between these tools thus we perform comparative analysis of motion data from two alternate human motion-capture systems (high-resolution Vicon vs low-resolution Kinect). In addition to traditional resolution/accuracy study, data for multiple trials of motions were captured and examined to verify motion capture fidelity and the role of pre- and post-processing (calibration and estimation). In our work, we adapt Principal Component Analysis (PCA) approaches and K-Nearest Neighbors (K-NN) method for subject classification.

Knee bracing has been used to realize a variety of functional outcomes in both sport and rehabilitation application. Traditionally, the design of exoskeletons (from choice of configuration to selection of parameters) as well as the process of fitting this exoskeleton (to the individual
user/patient) has largely depended on intuition and/or practical experience of a designer/physiotherapist. However, improper exoskeleton design and/or incorrect fitting can cause buildup of significant residual forces/torques (both at joint and fixation site). Performance can be further compromised by the innate complexity of human motions and need to accommodate the immense individual variability (in terms of patient–anthropometrics, motion–envelopes and musculoskeletal–strength). In our work, we propose a systematic and quantitative methodology to evaluate various alternate exoskeleton designs using kinetostatic design optimization and twist-/wrench- based modeling and analysis. This process is applied in the context of a case-study for developing optimal configuration and fixation of a knee brace/exoskeleton. An optimized knee brace is prototyped using 3D printing and physically tested.

Recent research on exoskeletons has examined ways of improving flexibility, wearability as well as reducing overall weight. Very few exoskeletal systems, however, have succeeded in satisfying all these criteria due to the complexities engaged in human joint motions and loading. Compliant mechanisms offer a class of articulated multibody systems that allow relatively stiff but lightweight solutions for exoskeleton/braces. In our study, we introduce Parallel Coupled Compliant plate (PCCP) mechanism and Pennate Elastic Band (PEB) spring architecture and evaluate them. PCCP/PEB system provides both flexibility and extreme stiffness to user with respect to posture/angle of knee joint. The performance of PCCP/PEB system was verified by 3D printed physical exoskeleton prototype.

The overall human subject measurement and adjustable smart brace controller are integrated within a Matlab based acquisition, analysis and control framework. Motion measured by low-cost devices (Kinect and Wii Balance Board) was used to calculate load at knee joint then the
smart knee brace automatically adjusted parameters of brace to control load at the knee joint based on prescription by therapist or doctor.
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1 Motivation

The Smart Health Paradigm has opened up immense possibilities for designing Cyber Physical System (CPS) with integrated sensing and analysis for data-enhanced health care decision making and conduct. Specifically, we seek to improve our understanding of motor rehabilitation via intelligent robotic exoskeletons/braces. This, in turn, entails improving our understanding of the complex interplay of the human neuro-musculo-skeletal system with its environment, the bidirectional exchange of power (motions and forces) and decision-making and implementation. The ultimate goal is to intelligently customize and interactively modulate the associated dynamical human behaviors.

To accomplish this goal, we propose to develop a comprehensive cyber-physical framework of quantitatively sensing, analyzing, decomposing, understanding, and modulating human behaviors through wearable robotic systems and low-cost motion capture devices for home-based progressive rehabilitation. We will concretize this framework by examining individualized rehabilitation with an intelligent, articulated, and adjustable lower limb orthotic brace used to manage a major public health issue, knee osteoarthritis (OA).

Figure 1. Epidemic of Arthritis (a) Cover of Time magazine (b) Projected Prevalence of Doctor-Diagnosed Arthritis Among U.S. Adults Ages 18+ Years, 2005-2030 [1]
Osteo-arthritis (OA) is a progressive degenerative disease afflicting a large and growing number of Americans [2]. OA is predicted to become the fourth leading cause of disability as the population ages and becomes more obese. OA induced pain and knee instability limit function and slow or interrupt locomotion tasks [3]. An estimated 24 - 37% of arthritic adults in the US report reduced capacity for “normal daily activity”, which predisposes them to further health decline associated with inactivity. Thus, earlier onset in increased patient populations, chronic and prolonged course, and costly treatments make osteoarthritis an important public health issue [3, 4]. According to research by Hootman et al. [1], by the year 2030, the prevalence of clinically-diagnosed arthritis is estimated 67 million (25% of the projected total adult population) adults in US, as shown in Figure 1-(b).

![Figure 1.](image1.jpg)

(a) Joint realignment for OA and (b) OA knee brace [5]

By 2015 the number of knee replacements and joint realignment surgery (osteotomy) as shown in Figure 2, are expected to 1.3 million and cost $49 billion [6]. Both procedures are expensive, are variably successful, require long recoveries, and risk surgical morbidity [7]. However, OA bracing (see Figure 2) is a conservative (non-operative) intervention that has been shown to mediate pain, and improve function and quality of life. OA braces are thought to mechanically unload the medial compartment by applying external forces to strengthen the knee [8], thereby slowing progression of the disease.
The human knee is a complex-joint that exhibits six degree-of-freedom (DOF) motion between the femur and tibia as depicted in Figure 3. The ensuing knee displacements and rotations (and their time-derivatives), and forces/moments, need to be resisted by a complex-combination of muscular-, ligamental-, and meniscocapsular- constraints.

The knee joint is highly loaded during daily life. Kutzner et al. [9] measured the contact forces and moments acting on the tibial component using an instrumented knee implant. Average peak resultant forces (in percent of body weight) were repeatedly 346% for stair descending, 241% for level walking and 246% for erect standing.

![Figure 3. Nomenclature for knee motion degree of freedom](image)

![Figure 4. (a) Knee flexion angle of OA patient during level walking [10] and (b) Mean absolute scores of pain domain of the WOMAC (Worst score is 500 mm) [11]](image)
Peak shear forces were about 10~20% times smaller than the axial force and the range of moments in the frontal plane was 1.6% to 2.91% (in percent of body weight moments) throughout all daily activities.

The gait pattern of OA patient was studied by Kaufman et al., who found an initial loading response and smaller flexion angle during swing as shown in Figure 4 [10]. The shaded region represents the normal range and thick solid line shows knee kinematic pattern of the patient with OA. Kirkley et al. showed a significant improvement in i) disease-specific quality of life and ii) function with bracing as shown in Figure 4 [11].

Immense variability exists across a population based on gender, and age, which is compounded by individual differences stemming from level of conditioning and stage of disease and/or therapy. Hence, there is great interest in carefully personalizing entire rehabilitation programs, both in terms of user-device ergonomics and regimen parameters, to enhance patient outcomes.

However, a sound understanding of interplay between such design and control of such articulated bracing with individual neuro-musculo-skeletal deficits for realizing desired dynamic interaction patterns is lacking. Moreover, such a rehabilitation process tends to be labor-intensive, relying on inpatient diagnostic and therapeutic procedures that are administered by a clinician working with a single patient at a time. As the number of patients increase, limited availability of rehabilitation centers with specialized equipment/personnel support offer serious constraints, which is particularly acute for those people living in rural or remote locations. In order to address these shortcomings, we undertook a visual-sensing/dynamical-systems perspective focusing on 4 stages.
• **Visual Sensing:** Track and record, under carefully controlled conditions, low-level dynamic behaviors and user/device interaction in the homes with ongoing internet-based monitoring and support through commercial-off-the-shelf (COTS) devices, e.g. Kinect sensor, Wii Balance Board;

• **Data Representation:** Abstract the visual dynamics and interactions at the intermediate scale into parametric and composable low-dimensional manifold representations;

• **Diagnostic Assessment:** Link visual representations to quantitative biomechanical assessment of the individual patients to aid the development of individualized user model and exercise regimen; and

• **Inference Implementation:** Virtually or remotely assist in the progressive parametric refinement of exercises and adjustment of articulated OA bracing device.

In this process, we seek to systematically analyze the role and contributions due to each stage both from a fundamental (science) as well as the implementation (engineering) perspective.

### 1.1 Home-based Rehabilitation Scenarios

![Figure 5. Advantage of home-based rehabilitation system](image_url)
Clinical rehabilitation has long relied on quantitative motion capture coupled with subsequent computational analysis to help with diagnosis and treatment of various movement disorders [2]. While surgical as well as more conservative (non-operative) interventions are available to mediate pain and improve function and quality of life for chronic cases, the systematic performance of a structured exercise regimen can alleviate and slow the progression of the disease. However, immense variability exists across a population based on gender, and age, which is compounded by individual differences stemming from level of conditioning and stage of disease and/or therapy. Hence, there is great interest in carefully personalizing entire rehabilitation programs, both in terms of exercise-regimen parameters, to enhance patient outcomes.

<table>
<thead>
<tr>
<th>Author</th>
<th>Intervention</th>
<th>Period</th>
<th>Sample</th>
<th>Main Results</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anderson et al.</td>
<td>Hospital vs. home-based rehabilitation (RCT)</td>
<td>6 months</td>
<td>86</td>
<td>No difference in outcomes, lower costs in rehabilitation</td>
</tr>
<tr>
<td>[12, 13]</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Byford et al.</td>
<td>Short-term family placement scheme</td>
<td>3 months</td>
<td>120</td>
<td>Increased functional outcome, decreased cost</td>
</tr>
<tr>
<td>[14]</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Gladman et al.</td>
<td>Domiciliary vs. hospital based rehabilitation</td>
<td>6 months</td>
<td>327</td>
<td>No difference in outcomes, cost in domiciliary service</td>
</tr>
<tr>
<td>[15]</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hui et al.</td>
<td>Day hospital vs. conventional care (RCT)</td>
<td>6 months</td>
<td>120</td>
<td>No difference in functional outcomes, no difference in cost</td>
</tr>
<tr>
<td>[16]</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Young &amp; Forster</td>
<td>Home physiotherapy vs. day hospital</td>
<td>8 weeks</td>
<td>95</td>
<td>No difference in functional outcomes, decreased cost in home physiotherapy</td>
</tr>
<tr>
<td>[17]</td>
<td></td>
<td></td>
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</tr>
</tbody>
</table>

Table 1. Comparative study in rehabilitation (RCT=randomized controlled trial)

While inpatient therapy remains the preferred means for therapy (in terms of recovery times), home-based rehabilitation programs have gained importance and relevance due to the considerable flexibility afforded in tailoring the scheduling, intensity and duration of the rehabilitation regimen. Considerable literature has noted the benefits of home-based rehabilitation as a viable approach to provide treatment for patients [18].
Numerous studies have shown that a program of continuing self-directed exercises for patients discharged home, supervised once a week by therapists, was as effective as outpatient day hospital therapy (and definitely more resource efficient). Table 1 highlights some of the comparative studies carried out between home-based and day hospital/outpatient therapy which also support the assumption that home-based rehabilitation might be more economical, with comparable effects in terms of functional gains.

At the same time an improved fundamental understanding of OA rehabilitation and technological progress has opened new vistas in the delivery of such rehabilitation regimen. The digital revolution of the past several decades has promoted the trend towards quantitative monitoring by using computerized data-acquisition technologies, with the fields of rehabilitation being no exception.

The principal benefits of such systems stem from their ability to quantitatively and automatically monitor (record/replay) various physical performance characteristics and most importantly to post-process (analyze/compare) this data. Recent therapeutic modalities, such as constraint-induced therapy, have been shown to significantly faster restorative function [19]. Studies have shown that “constraint-induced therapy” can improve movement ability with intensive, supervised training [19, 20].

It must be noted, labor-intensive neuro-rehabilitation procedure is a primarily mediated by robotics, with development and deployment of robot-assisted-therapy devices (‘rehabilitators’) that physically-interact with patients in order to assist in movement therapy [21-24]. Such robotic devices augment the therapist by guiding the patient through intensive, repetitive practice of functional movement, with several studies documenting their successes [25-27]. However, the
significant costs and other logistics associated with using such devices make it possible to deploy these only in hospital settings.

Thus, it is in the final stages of translating these advances to the home-based rehabilitation setting that things have faltered, principally due to the lack of readily available specialized equipment affordable by each OA patient. For example, the recommended home-based exercises for OA patients, who are (mildly) affected by OA, are principally unassisted exercises prescription. However, the lack of structured and monitored performance of the exercise (by a therapist or machine) can mitigate the achievable benefits as they may not appropriately stress the affected limbs/joints/muscles.

![Figure 6. Examples of home-based rehabilitation system by using (a) Kinect [28] and (b) Wii Balance Board [29]](image)

Advances in miniaturization of processors/sensors/actuators have created a new generation of smart embedded product that offer significant and yet cost-effective functionality and performance. For example, numerous commercial-off-the-shelf computer-interface-devices e.g. Kinect, Wii balance board and Novint Falcon that are intended primarily for gaming applications, not only sense a person’s motions but also apply forces during such motions. Similar to existing robotic therapy, these devices serve as interfaces that stimulate movement profiles of movement, as well as create customizable patterns of active/passive motion and force assists to user motions.
Thus, we believe that there is a class of problems where low cost commercial-off-the-shelf devices, coupled with rehabilitation therapy protocols, opens up the possibility of widespread deployment as a truly inexpensive home-based trainer.

Figure 6 shows recent/emerging examples of home-based rehabilitation systems have integrated with commercial-off-the-shelf devices. The performance of current home-based applications is limited to assisting or guiding patients’ motion. Although they can provide quantitative analysis results, the quality of data is mostly limited by the resolution of measuring system. We will explore these various aspects in developing a comprehensive rehabilitation framework for alleviating pain and enhancing performance in knee deficient patients.

1.2 Proposed Cyber Physical Framework for Home-based Rehabilitation

The overall cyber-physical framework consists of a i) Patient Interface (ultimately intended to be home-based) and ii) Therapist Interface (intended to be at a remote central hospital location) that are connected through the Internet (Figure 7).

The emphasis on modularity and bi-directional parametric coupling in all aspects of development of this framework is intended to facilitate “plug-n-play” functionality and to achieve distributed implementation on different computational platforms. As illustrated in the physical (left) module of Figure 7, a knee OA patient at home interacts with a patient interface using the home-based COTS devices (Kinect sensor, Wii Balance Board) and OA brace, which serves as a manipulandum to quantitatively capture the patient motion, activity characteristics, and deploy customized exercises.
By leveraging internet-based networking, such a patient interface could then be connected with a therapist interface at a knee OA rehabilitation center. The remote rehabilitation therapist would then be able to monitor this patient’s sensorimotor performance.

We focus on the conservative, and more economically feasible intervention by developing an intelligent mechanical OA braces designed to not only passively constraint motion, but also to redirect and reuse energy at various stages of the gait cycle. The OA brace is actively adjustable and articulated, incorporating external in-parallel elastic members, which modulate transfer of stored spring energy to limb-segment potential energy at appropriate interval of the gait cycle. Therefore, this low-cost, home-based hardware coupled with a customizable regimen of motion-based limb movement therapy has tremendous potential to make telerehabilitation services a viable option. However, a validated framework is necessary to facilitate both the quantitative
diagnostic monitoring and individualized motor rehabilitation of limb dysfunction as well as be well suited for deployment in non-institutional settings.

We focused on developing new methodologies to fill in the gap between quantitative measurements and clinical diagnostic measures – these form the software component to drive interactions. Quantitative biomechanical assessment was achieved by musculoskeletal modeling using corresponding brace and motion capture data. This is intended to aid the development of a customized user model and assist in the progressive parametric refinement of exercise and adjustment of articulated OA braces.

We begin with a brief overview of the diverse topics ranging from: i) motion capture technology, ii) motion analysis, iii) commercial off-the-shelf computer gaming interfaces and iv) knee exoskeletons. The discussion in each case is necessarily brief and is intended to set the stage for the overall work. A more in-depth literature is found within each subsequent chapter.

1.3 Overview of Rest of The Dissertation

Chapter 2 serves as the background of our research – with emphasis on current motion capture/analysis technology and home-based smart health paradigm - being introduced. In Chapter 3, the two motion capture systems – a low-end Kinect sensor and high-end Vicon system – were examined with respect to quantitative lower limb motion estimation, and with suitable post processing their potential for applicability for clinically relevant use. This work was published in 2013 IEEE International Conference on Automation Science and Engineering and 2013 Digital Human Modeling Symposium [30, 31].
Chapter 4 and 5 examined the use of kinetostatic and screw-theoretic analysis tools to provide a systematic and quantitative framework for design, formulation and evaluation of knee exoskeleton. Specifically we emphasize on automatic and systematic design process which can be easily extended to upper limb or whole body exoskeleton. This work was published in 2013/2014 ASME International Design & Engineering Technical Conferences (IDETC) [32-34].

In Chapter 6, a novel compliant mechanism with nonlinear spring (PCCP/PEB mechanism) is introduced with the result from a full-scale physical prototype presented. This work was published as an abstracted in IDETC/CIE 2014 [34] and the manuscript is accepted with ASME Journal of Mechanism and Robotics [35].
In Chapter 7, a prototype of the home-based rehabilitation system that consists of measuring system, smart knee brace and host system is presented. Measuring system comprises the Kinect and Wii Balance Board to measure both kinematic and static force data from the patient. The smart knee brace communicates with host system and provides customized force and torque (as measured and calculated by low-cost sensors) while being worn by the patient, as shown in Figure 9.
2 Background

We present the foundations upon which we will build our proposed solutions to the identified research needs. These foundations encompass a necessarily brief discussion of the following critical aspects to telerehabilitation approaches, immersive therapeutic environments, the role of quantitative assessment in biomechanical modeling, identification and user-specific customization and critical trends in rehabilitation methodologies.

2.1 State of the Art Human Motion Capture

![Motion capture lab](image)  
![Vicon optical tracking system](image)  
![Kistler force platform](image)

Figure 10. (a) Motion capture lab, (b) Vicon optical tracking system [36] and (c) Kistler force platform [37]

Motion capture is a technique where high-speed digital cameras are used to track retro-reflective markers placed over body segments (head–neck, trunk, pelvis, arms, forearms, thighs and feet), from which 3D human movement is inferred using reconstructed 3D marker trajectories. As depicted in Figure 10, motion capture system usually consists of multiple cameras which are integrated with force platforms embedded in the floor or stairs to record joint angles and ground reaction forces while persons ambulate over stairs or level ground.

The Vicon motion capture system is an infrared optical marker tracking system that offers sub-millimeter resolution of 3D spatial displacements recording faster than 120Hz. Our system implementation consists of eight cameras (F20-MX) outfitted with IR optical filters and an array
of IR LEDs. The dots, arranged on the hand of the subject (human or non-human primate), reflect the IR radiation emitted by the LEDs [36]. Kistler force platform measures the six-dimensional components of force and torque applied to the surface and its point of application [37].

2.2 State of the Art in Human Motion Analysis

Clinicians who treat human movement pathologies examine not only access movement but also, simultaneous neuromuscular function both before and after treatment interventions. However, synthesizing detailed descriptions of the elements of the neuromusculoskeletal system with measurements of movement remains a major challenge. If successful, this would not only create an integrated understanding of normal movement but can also help establish a scientific basis for correcting abnormal movement. However it must also be noted that using experiments alone to understand movement dynamics has two fundamental limitations. First, important variables, including the forces generated by the muscles, are not generally measurable. Second, it is difficult to establish cause-effect relationships in complex dynamic systems from experimental data alone [38].

Dynamic simulation of movement that integrates anthropometric, anatomic and physiologic elements of the neuro-musculoskeletal system combined with multi-joint mechanics potentially provides such a framework. Muscle-driven dynamic simulations complement experimental approaches by providing estimates of muscle and joint forces, which are difficult to measure experimentally. Simulations may also identify cause-effect relationships allowing for “what if?” studies to be performed where for example, the excitation pattern of a muscle can be altered and the resulting motion can be observed.
Figure 11. Simulation result from (a) Anybody Modeling System, (b) OpenSim and (c) Visual3D

The AnyBody Modeling System™ [39] is a computational framework for simulating the human musculoskeletal performance, used to estimate individual muscle force, joint force and moment, metabolism, elastic energy and antagonistic muscle actions. An alternation is OpenSim [40], a free open-source platform for modeling, simulating, and analyzing the neuromusculoskeletal system. OpenSim, developed and maintained on Simtk.org, provides a platform on which a library of simulations are being tested, analyzed, and improved by the biomechanics community through multi institutional collaboration. The Visual3D [41] is another commercial analysis tool for biomechanical modeling and analysis, which is used to measure and quantify movement as collected by a 3D motion capture systems. Simulation from the three major human motion analysis tools are shown in Figure 11.

2.3 Commercial off the Shelf Computer Gaming Interfaces

However it is in the final stages of translating these advances to the home-based rehabilitation arena that things have faltered, principally due to the lack of readily affordable specialized equipment. For example, recommended home-based exercises while being monitored for mildly affect OA patients are principally unassisted exercises. However, the lack of exercise
while being monitored for correct movement by the therapist can mitigate the achievable benefits as they may not appropriately stress the affected limbs, segments, joints and muscles. In recent years, commercial-off-the-shelf computer-interface-devices, e.g. Kinect and Wii Balance Board, now offer a low-cost real-time means for monitoring 3D motion-performance of users. Thus, we believe that there is a class of pathologies where low cost commercial-off-the-shelf devices, coupled with rehabilitation therapy protocols, opens up the possibility of widespread deployment as a truly inexpensive home-based treatment modality.

![Kinect and Wii Balance Board](image)

*Figure 12. (a) Kinect by Microsoft [42] and (b) Wii Balance Board by Nintendo [43]*

The advent of the Kinect system potentially offers an opportunity to track human motion in real-time at a fraction of the cost of conventionally-equipped gait labs. Together with computational human modeling and analysis tools, this offers an opportunity to potentially gain insight into functional performance of humans outside the limiting motion-capture- lab setting. However, multiple factors need to be considered in supporting the deployment of such a framework. First there exist significant differences in the capabilities and ease-of-use between these tools, necessitating a careful evaluation. The Kinect system consists of Kinect sensor interfaced with either the Kinect Windows API or the Kinect Windows SDK. The Kinect system streams color image, depth image data and can recognize and track human motion up to 30Hz in the frontal plane. The Kinect can recognize up to six human objects in the field of view and up to two objects can be tracked in detail. Furthermore information is presented in absolute positions
relative to the Kinect camera coordinate, and with orientation of segments expressed in quaternions and rotation matrices [42].

Although similar in structure as a force platform, the Nintendo Wii Balance Board (WBB) is primarily a gaming device that measures center of gravity and weight of subject. Although developed for entertainment, it has the potential to be an inexpensive and readily accessible force platform for home-based rehabilitation. Bartlett et al. measured accuracy and repeatability of the WBB. Individual WBB weight sensor has an accuracy of ±0.61 kg and repeatability error of ±0.52 kg [44]. WBB is one of peripheral devices for Wii fit video game. This device has similar structure with a force platform for motion capture laboratory but they are different in resolution, accuracy and degree-of-freedom. One of major differences between Wii Balance Board and force platform is degree-of-freedom (DOF) of force transducer located at four corners and overall system. For force platform, each force transducer has 3-DOF but force sensor of Wii Balance Board can measure the force in only one direction. Also force platform measures forces and moment on top plate in 6-DOF, but the Wii Balance Board measures perpendicular force and moments in two directions, as shown in Figure 12.

![Figure 13. FBD of (a) force platform and (b) Wii Balance Board](image-url)
2.4 Current Knee Brace and Exoskeleton Design

Knee braces, as shown in Figure 14, are used to partially compensate for knee motion/forces in order to protect and stabilize the knee during regular activities. The ideal knee brace should be designed to: i) not interfere with normal knee function, ii) decrease risk of injury to other joints/limbs, iii) to producing synergistic motion-force constraints together with physiological components. For sport/rehabilitation, the emphasis is on lightweight, passive kinematic braces that primarily serve to constrain selected kinematic motion (but typically do not contribute to altering the force distribution).

![Figure 14. (a) Knee brace by OSSUR [45], (b) Quasi-Passive Knee Exoskeleton by Yale GRAB Lab [46], (c) BLEEX System by Berkeley Robotics & Human Engineering Lab [47] and (d) HULC System by Lockheed Martin [48]](image)

In parallel, there are efforts at developing robotic-exoskeletons to serve as lower limb augmentation systems. In emphasizing active-power (motion and force) interactions, they often require significant infrastructure, i.e. a mobile supporting frame, added actuation, embeddable electronics and high-capacity mobile power-sources. Typically, this active-power interaction modality translates into somewhat bulky and unwieldy designs, as evident in Figure 14. While previous efforts have demonstrated progress and enhanced our knowledge, such efforts have not approached the compactness, efficiency or ergonomics required of fieldable systems.
In general, the design-specifications for the knee-brace vary based upon the task. It can become quite a challenge to even correctly identify the right set of specifications to which the knee-brace is to be designed. In our current work, we choose a modest starting point by addressing the development of an adjustable passive torque-assist knee-brace, focusing on knee flexion/extension in the sagittal plane, to assist with squatting motions. Addressing multiple degrees-of freedom will be critical for real-world deployments of smart knee-braces and remains an active research topic. However, this process is clearly much more involved for multi-DOF joints (such as complete knee or ankle motions). Hence, we will remain focused on development of an adjustable torque-assist knee-brace for flexion/extension motions alone.

In human locomotion, the joint mechanical impedance is continuously modulated either voluntarily or reflectively to achieve locomotive function. Lee and Hogan [49] measured the mechanical impedance of ankle during level walking using 2nd order time-varying model and lower-limb exoskeleton. They found that viscosity and stiffness of the ankle are varied with respect to the phase of motion. Aguirre-Ollinger et al. [50] controlled impedance of leg by applying negative inertia using one-degree-of-freedom exoskeleton. Design goals of exoskeleton: the exoskeleton’s mechanism as transparent to the user as possible, and provide the assistive forces, were pursued to achieve agility and functional goal of system. In our work, given relatively slow motions, torsional-joint-stiffness is the static approximation to the dynamic torsional-joint-impedance and can be calculated as the partial derivative of the torque-assist with respect to the knee angle.
2.5 Quantitative Biomechanical Assessment and Computational Modeling

Quantitative assessment and the automated monitoring process hold considerable promise not only for enhancing the quality of individualization possible, but it may also effectively decouple the problem of diagnosis and prescription from aspects of the delivery. In particular, three primary categories of motor patterns are used for assessing performance of movement tasks: i) joint kinematics describe motion between inter-linking body segments when performing a task; ii) joint kinetics provide a measures of the net causative demand placed on joint actuators (protagonist muscles) to perform a movement; and iii) neuromuscular activation profiles (EMG) provide insight into the coordinated muscle activity when performing a movement task.

The nuances of acquiring these three types of biomechanical measures in an accurate, valid and reliable way are well documented [58], which are time-dependent and require some kind of frequency or time normalization procedures before averaging, or statistical processing before compared to pathological movement patterns [59-61]. Bilateral comparisons are particularly powerful for assessment and can be reduced to a single measure such as the symmetry index [62]. In addition to direct measures from the neurons, muscles, and limbs, considerable effort in recent years have focused on developing computational models of the human neuro-musculoskeletal system.

Developing such computational models is challenging because of the intrinsic complexity of modeling human motion generation, entailing the merger of cognitive, neural, skeletal and muscular subsystems, which must be represented accurately in order to provide insights into musculoskeletal performance. Existing parametric computational models ranges from artificial neural network-based models that relate upper limb movement to muscle EMG signals, to full-
fledged musculoskeletal modeling and analysis software which include OpenSim or Anybody [39, 40]. Generic models, using these software systems, however need to be customized by adjusting numerous parameters to match those of the specific individual, a process which can be tedious.

However, given a model structure, considerable literature exists for parameter identification, especially online adaptive parameter estimation. In our efforts we focus on adapting these techniques to aid the creating of customized patient models.

2.6 Quantitative Progressive Exercise Rehabilitation Regimen

Most rehabilitation protocols are individualized and of a progressive nature to some extent, with appropriate, valid and reliable quantitative techniques. One such broad-based rehabilitative paradigm called Quantitative Progressive Exercise Rehabilitation (QPER), is based on the quantitative measure of physiologic and functional deficits of the individual [63, 64] while addressing some of the limitations of standard rehabilitation strategies.

Specifically by factoring the physiological changes that occur with development and/or aging, the pathophysiologic changes that occur as a result of dystonia or disability together with the appropriate quantitative physiologic and functional deficits measurements, the most suitable treatments for the individual in terms of the individual’s indications and contraindications for exercise, including appropriate mode, intensity, frequency and duration of exercise, based on exercise physiology principles. Additional factors include immersion designed to encourage high rates of compliance and adherence; ease of program implementation; and overall cost-effectiveness.

Such QPER strategy has been successfully employed in the quantitative evaluation and treatment of several disability groups, including frail [65-67] and well elderly [66-68], those with
osteoarthritis [63-65, 69], juvenile arthritis [70] and patients with multiple sclerosis[71]. In this work we focus on adaptation of QPER strategy into a form that will allow remote prescription of individualized and progressive exercise rehabilitation programs.

2.7 Immersive and Interactive Therapeutic Environments.

In the past decade, numerous interactive therapeutic environments have been tested and validated. In essence such environments permit performance controlled sensorimotor experiments with quantitative capture, storage and playback of user interaction either when exposed to active or passive stimuli. Several researchers [3, 53] examined the application of robotic devices and automation technology to assist, enhance, quantify and document neuro-rehabilitation. One important benefit includes the ability to install the therapeutic systems in user homes and remotely monitor their progress via the internet, which we will also pursue.

However, devices tend to be specialized/custom-built devices which limit ubiquitous access. As a result, researchers have begun to examine the use of truly low cost, mass-produced commercial-off-the-shelf force-feedback devices (commonly for gaming applications) for rehabilitation therapy applications. A modified commercial-off-the-shelf force-feedback joystick (Microsoft Sidewinder), with an arm support was coupled to a target tracking scenario to serve as an exercise protocol, implemented as a downloadable web-based, Java applet game. While emphasis was on examining the use of artificial assistive forces (generated via the force feedback joystick) on mediating arm movements, the quantitative measurement capabilities to facilitate diagnosis was not explored, the way we intend to develop.
2.8 Telerehabilitation

In recent years, by leveraging the power of internet, real-time video transmission has come to supplement audio and data transmission for telerehabilitation applications [51]. In a representative scenario, the clinic-based tele-rehabilitation specialist observes the remote-clients tasks, and subsequently instructs/coach the home-based team. Real-time video teleconferencing provides a visual and explicit exchange of information during this process with significant consumer (both patient and therapist) acceptance. However multiple camera views may be necessary for therapist to recognize patients’ activity patterns, thereby increasing the infrastructure (cameras, network-bandwidth) requirements of such a tele-counseling approach.

To date, quantitative assessment is difficult with inexpensive videoconferencing systems, which are unable to leverage the quantitative computational infrastructure to assist the assessment.

Many research groups are therefore beginning to consider augmenting video information with collected quantitative physiological information [52]. Although such approaches have been intended principally for cardiac, respiratory and diabetes management and they are not being explored in the telerehabilitation context, which is under consideration in this work.

2.9 User-Specific Customization

There is a considerable variation of performance, function and disability across individuals in a population and hence rehabilitative devices and services necessarily have to be one-of-a-kind and subject specific design [54]. However, traditional user-customization models have had limited success. On one hand, categorizing individuals into one of many “stock sizes” (on the basis of limited physiological measurements) and using “off-the-rack” products (with a few adjustable features) results in poor fit and diminished performance.
On the other hand, adopting a “tailor-made” approach for enhanced user-customization comes with the significant expense for skilled manual-labor and an iterative fitting process [55]. In light of this, new paradigms for rapid development of inexpensive, high quality rehabilitation aids customized to the specific needs of the individual patients have emerged. Over the past decade [56, 57]. As such, implementing an automated process of deriving user-specified and iterative design refinement with concomitant speedup and cost reductions that facilitates delivery of rehabilitation services is one of the goals of this work.
3 Motion Measurement

Because performance, function and disability across individuals in a population, new paradigms for rapid development of inexpensive, high quality rehabilitation aids customized to the specific needs of the individual patient. The quantitative assessment and automation of monitoring hold considerable promise not only for enhancing the quality of individualization possible, but also effectively decoupling the problem of diagnosis and prescription from aspects of the delivery.

In order to retain clinical relevance, we focused attention on the quantitative evaluation of squatting, the “bending of the knee so that buttocks rest on or near the heels”. Because work-related musculoskeletal disorders are one of the greatest occupational health concerns and stooping/squatting is the dominant cause of injury. The squatting posture is one of most prevalent in agriculture, construction, mining and other profession not usually considered to be physically demanding so it is important to understand and recognize the inherent risks [72].

![Figure 15](image)

Figure 15. (a) Squatting motion [73], (b) Coordinate of joints and definition of joint angle and (c) Knee flexion angles for the preoperative and postoperative and control groups
The Functional Movement Screening (FMS) test is a method of evaluating basic movement abilities. The test is comprised of seven fundamental movement patterns including squatting that require a balance of mobility and stability. Deep squatting is used to assess bilateral, symmetrical and functional mobility between the hips, knees and ankles as shown in Figure 15 – (a) [73].

From a clinical perspective, the degree of flexion is a critical outcome measure in the differential diagnosis of knee flexor deficits as well as monitoring improvement owing to treatment. Ramsey et al. [74] introduced the effect of anatomic realignment on muscle function. They show that static alignment improves medial laxity, stability and adduction moment but requires accurate measurement of knee angle. Figure 15 - (c) reproduces knee flexion angles for the pre-operative, post-operative and control groups, originally presented in Ramsey et al. Hence, we focus on quantitatively comparing the knee joint angle and coordinate of subjects.

This chapter consists of 1) motion capture with optical tracking system and Kinect, 2) calibration and estimation, 3) human identification, 4) static measurement by Wii Balance Board and 5) knee performance measurement.

### 3.1 Motion Capture with Optical Tracking System and Kinect

#### 3.1.1 Synchronization (Vicon and Kinect)

![Figure 16. Motion capture setup by synchronized Kinect and Vicon system](image)
To begin, motion-capture of squatting was recorded using two alternate synchronized systems – a high-fidelity Vicon MX-F20 System [36] together with a low-cost/resolution commercial-off-the-shelf Kinect sensor [42]. The Vicon motion capture system track retro-reflective markers placed over body segments (head–neck, trunk, pelvis, arms, forearms, thighs and feet), from which 3D human movement was inferred using reconstructed 3D marker trajectories. The Vicon MX system was synchronized with the Kinect system via a video synchronizer (Kistler 5610), as seen in Figure 16, to trigger and record simultaneously.

The Kinect system consists of Kinect sensor interfaced via the Kinect Windows API or the Kinect Windows SDK. The Kinect system streams color image, depth image and also recognizes and track human skeleton joints. Furthermore spatial information is presented in absolute positions relative to the Kinect camera coordinate system.

![Marker set for Vicon system and skeletal tracking model of Kinect](image)

**Figure 17.** (a) Marker set for Vicon system and (b) skeletal tracking model of Kinect [42]

### 3.1.2 Limitation of Kinect System

Kinect system’s skeletal tracking within recognize users facing the sensor and Kinect system recognizes person within 0.8 and 4.0 meter from the camera, suggesting a practical range of 1.2 to 3.5 meter. We observed the performance of Kinect system depends critically on the location of
subject within the workspace. As shown as Figure 18, detection of second subject causes the tracking performance of the primary subject to deteriorate. Thus, although it is able to simultaneously track multiple human subjects: i) recordings were restricted to one subject at a time; ii) all body segments were within the workspace. Figure 18 - (a) shows the effect of second subject detection, (b) and (c) depict the relationship between subject’s position in the workspace and Kinect performance, while (d) illustrates the ideal subject location for motion capture and corresponding inferred knee joint angles.

Dutta assessed the accuracy of Kinect system by using Vicon system as a gold standard. The Kinect system can detect 3-D coordinates of markers with RMS error of 5.7mm to 10.9mm over the range of 1.0m to 3.0m [75]. Gabel et al. in examining the capability of Kinect to extract gait information [76]. Pedro and Caurin showed that Kinect system has good repeatability in its central region and it progressively worsens with increasing depth [77]. Data from Kinect and optical tracking system was compared and it was concluded that Kinect has significant potential as avlow-cost alternative for real-time motion capture system [78, 79]. Sinthanayothin et al. surveyed different techniques for vision-based human motion captures and analysis along with the capability of Kinect system for human skeleton tracking [80]. Rincón et al. studied the feasibility of recovering human pose with data from a single camera by using particle- or Kalman-filters [81]. However, most previous work focused on directly comparing motion-capture performance between the Kinect system and high-end measurement systems (Vicon); improving motion-capture capabilities without considering objective performance assessments; and especially without clear consideration of resolution enhancement.
In our study, fifteen healthy male subjects without knee injury history were studied and marker set for Anybody Modeling System (AMS) full body gait model are adapted. Reflective markers were fixed on 41 anatomical landmarks as presented in Figure 16 and subjects were asked to perform a series of gentle squatting trials and each subject was given several practice trial before recording. Five measurement trials were recorded for each movement task as well as a standing reference trial before the movement trials.

3.1.3 Lower Limb Kinematics and Dynamics

Tracking markers were placed on the distal and proximal end of each segment which provide a marker reference frame and skeletal landmarks generate anatomical reference frame as shown in Figure 19. To analyze three dimensional motions of body segment, there must be at least three independent markers per segment. The markers on each segment must not be collinear, not in the same line and form a plane in space. Joint centers and the direction of axes are crucial.
information for biomechanics study and can be derived from spatial coordinate of anatomical landmarks.

**Tracking Marker Plane**
- Choose ‘TIB’ as origin
- $Z_m$ – line from ‘TIB’ to ‘LKNEE’
- $X_m$ – orthogonal to the tracking marker plane (TIB-LKNEE-LMAL)
- $Y_m$ – orthogonal to $Z_m \times X_m$

**Anatomical Plane**
- $Y_a$ – connect midpoint of ankle (LMAL-MMAL) and knee (LKNEE-MKNEE)
- $X_a$ – Perpendicular to the plane ($Y_a$ and MMAL to LMAL)
- $Z_a$ – orthogonal to $X_a \times Y_a$
- Choose ‘0.433 x ankle center to knee center’ as origin (center of mass)

In most clinical gait lab, it is impossible to track all anatomical landmarks in real time (especially for medial markers) because of blind spot of camera and interference of markers during subject’s motion. Hence, the calibration process is used to find the relationship between the marker reference frame and anatomical reference frame. Where the calibration markers are solely used to define the anatomical points from which the segment’s anatomical reference frame is defined. In our work, two medial markers (RMKNEE, RMMAL) were used to define the anatomical landmarks then removed after standing calibration.
Three tracking markers, for example RLKNEE, RLMAL and RTIB for the lower limb, define a tracking marker plane. The line from RTIB to RLKNEE defines as z-axis, y-axis is normal to tracking plan and x-axis is orthogonal to y- and z- axis of maker reference frame. For anatomical reference frame, Y-axis of is defined by connecting the midpoints of RLMAL-RMMAL and RLKNEE-RMKNEE, x-axis is perpendicular to the plan defined by y-axis/RMMAL/RLMAL and z-axis is orthogonal to x- and y- axis [58].

The matrix representation of a general linear transformation is transformed from one frame to another using a similarity transformation. As shown in Figure 19, $A_M^F$ is transformation from global reference frame to marker frame and $A_A^F$ is transformation from global reference frame to anatomical frame. Then the transformation from marker frame to anatomical frame $A_A^M$ can be represented as [82],

$$A_A^M = [A_M^F]^{-1}A_A^F$$

From calibration process with medial, we get position of medial markers (MKNEE, MMAL) with respect to marker frame.

$$P_{\text{medial},M} = \begin{bmatrix} x_{\text{mal},M} & x_{\text{knee},M} \\ y_{\text{mal},M} & y_{\text{knee},M} \\ z_{\text{mal},M} & z_{\text{knee},M} \end{bmatrix}$$

Then the coordinate of anatomical landmarks are calculated as

$$P_{\text{medial},F} = A_M^F A_A^M P_{\text{medial},M} \left( = A_M^F [A_M^F]^{-1}A_A^F P_{\text{medial},M} \right) \quad (1)$$

Figure 20 shows reconstructed anatomical landmarks and joint centers. The midpoint between the head of the fibula (LKNEE) and the medial epicondyle (MKNEE) of tibia and the midpoint between lateral and medial malleolus (LMAL, MMAL) are defined as knee joint and ankle joint respectively. The positions of lateral landmarks are same with the position of lateral
markers and medial landmarks are calculated by homogeneous transformation from marker frame to fixed frame as Equation 1.

![Image](image_url)

**Figure 20.** Reconstructed anatomical landmarks (lower limb) for squatting when (a) sitting and (b) standing position implemented by Matlab

The center of mass (COM), which is the origin of anatomical reference frame on shank, is defined by anthropometric data [58] as following:

$$COM_{\text{Shank}} = \text{ankle center} + 0.576 \times (\text{Knee center} - \text{ankle center})$$

The reaction forces at the proximal end of the segment in global reference frame is

$$F_{px} - F_{dx} = ma_x$$

$$F_{py} - F_{dy} - mg = ma_y$$

$$F_{pz} - F_{dz} = ma_z$$

(2)
where $a$ is acceleration at COM and $F$ is reaction force at proximal and distal position.

Figure 21. FBD of right lower limb (frontal view)

To calculate proximal moment, we need to transform both proximal and distal reaction forces and distal moment into the anatomical reference frame. The rotational equations of motion are

$$I_x \alpha_x + (I_z - I_y) \omega_y \omega_z = F'_{zd} l_d + F'_{zp} l_p + M_{xp} - M_{xd}$$

$$I_y \alpha_y + (I_x - I_z) \omega_x \omega_z = M_{yp} - M_{yd} \tag{3}$$

$$I_z \alpha_z + (I_y - I_x) \omega_x \omega_y = -F'_{zd} l_d - F'_{xp} l_p + M_{zp} - M_{zd}$$

Where $\omega$ is angular velocity, $M$ is transformed moment, $I$ is moment of inertia, $F'$ is transformed reaction force and $l$ is distance from COM to joint which is available from anthropometric data.

3.1.4 Error Analysis

Measurement error can be classified into two categories – accuracy and precision. Accuracy is the difference between the measured and actual value, and is assessed in terms of bias linearity.
and stability. Precision is the variation when the same part is measured repeatedly with the same device, and is characterized in terms of repeatability and reproducibility. Figure 22 depicts the coordinate of knee joint center and knee joint angle for multiple squatting trials. Since identical squatting motions (exactly same as previous trials) cannot be realized, classical methods to assess repeatability were not directly employed in this study.

Figure 22. (a) Right knee joint angle, (b) y coordinate of knee joint center during multiple squatting motions

The knee joint angle is often extracted and used for comparisons in gait and posture studies. The measured error of Kinect system was calculated as the difference between measurement of Vicon and Kinect measurement. The average error and standard deviation between the two are presented in Figure 23. The maximum average measured errors was 80.5mm (position) and 14 degree (Joint angle), which are hard to be acceptable in human motion research.
Knee joint motion was measured by tracking reflective markers located on anatomical landmarks of subjects. The accuracy of this measurement is prone to error due to soft skin artifact. Benoit et al. who compared skin mounted reflective markers with bone-pin marker method. Reported average rotational errors of skin-marker measurement were up to 4.4 ° and translational errors were up to 13mm during gait [83]. Whereas the true value is unknown, in this study, the work by Benoit’s provides a rough guideline for the range of residual error that can be expected for this work.

![Figure 23](image)

**Figure 23.** (a) Average and (b) STD of Measured Error of joint angle (degree) and position (x,z,y) during squatting (blue-angle, sky blue-x, yellow – z, brown – y)

### 3.2 Calibration and Estimation

Mathematical models of dynamic systems usually include poorly known parameters. However, determining the best estimate of unknown parameters provides an optimal estimate of the system’s actual behavior. Any systematic method to solve the problems aforementioned is referred to as an estimation process [84].
3.2.1 Linear Least Squares Estimation

The linear least squares method is considered as primary tool for process modeling because of the effectiveness and completeness. Although poor extrapolation and sensitivity to outlier of linear least squares method, many processes in science and engineering are well-described by linear least squares method [85].

The estimated and measured y-values can be written in compact form as

\[ \tilde{y} = H\hat{x} + e, \quad \bar{y} = Hx + v, \quad \hat{y} = H\hat{x} \]  \hspace{1cm} (4)

Where \( x \) is true x-values, \( v \) is measurement error, \( \tilde{y} \) is measured y-values and \( \hat{y} \) is estimated y-values.

The particular \( \hat{x} \) that minimize the sum square of the residual errors, is given by

\[ J = \frac{1}{2} e^T e \]  \hspace{1cm} (5)

Substituting equation (4) into (5),

\[ J = J(\hat{x}) = \frac{1}{2} (\tilde{y}^T \tilde{y} - 2\tilde{y}^T H\hat{x} + \hat{x}^T H^T H\hat{x}) \]  \hspace{1cm} (6)

Seek to find the \( \hat{x} \) that minimize, for a global minimum of the quadratic function of equation 6, we have the following requirements:

**Necessary condition**

\[ \nabla_x J \equiv \begin{bmatrix} \frac{\partial J}{\partial \hat{x}_1} \\ \vdots \\ \frac{\partial J}{\partial \hat{x}_n} \end{bmatrix} = H^T H\hat{x} - H^T \tilde{y} = 0 \]  \hspace{1cm} (7)

**Sufficient condition**
\[ \nabla^2 J \equiv \frac{\partial^2 J}{\partial \bar{x} \partial \bar{x}^T} = H^T H \]  

(8)

\( H^T H \) is always positive semi-definite. From the necessary condition, the normal equation is given by

\[ (H^T H)\bar{x} = H^T \bar{y} \]  

(9)

Because \( H^T H \) is always positive semi-definite and can be inverted to obtain the explicit solution for the optimal estimate:

\[ \bar{x} = (H^T H)^{-1} H^T \bar{y} \]  

(10)

### 3.2.2 Calibration and Estimation of Kinect data

Given all measurements were assumed to be available for simultaneous (batch) process, it was desirable to determine new estimates from previous measurements. The choice of basis functions usually depends on experience and knowledge of the particular relevant system [84]. As shown in Figure 24, a linear relationship between the knee center position as measured by Kinect and Vicon system was assumed, and therefore the X, Y, Z coordinates of knee center positions were considered linearly independent.

![Figure 24. (a) Translation of Kinect data and (b) Basis functions](image-url)
Although soft tissue artifact should be considered with optical tracking of human motion, given the required movement tasks were slow, controlled and constrained, the kinematic profiles derived from the Vicon motion capture system was considered as ground truth (assumption of zero model error) to which movement profiles from Kinect were compared [83].

\[
\tilde{y}_x = \begin{bmatrix} x_{\text{knee},1} \\ x_{\text{knee},2} \\ \vdots \\ x_{\text{knee},m-1} \\ x_{\text{knee},m} \end{bmatrix}_{\text{Vicon}} , \quad \tilde{y}_y = \begin{bmatrix} y_{\text{knee},1} \\ y_{\text{knee},2} \\ \vdots \\ y_{\text{knee},m-1} \\ y_{\text{knee},m} \end{bmatrix}_{\text{Vicon}} , \quad \tilde{y}_z = \begin{bmatrix} z_{\text{knee},1} \\ z_{\text{knee},2} \\ \vdots \\ z_{\text{knee},m-1} \\ z_{\text{knee},m} \end{bmatrix}_{\text{Vicon}}
\] (11)

From the necessary condition to minimize residual error, the optimal estimate is given by

\[
\hat{x} = (H^T H)^{-1} H^T \tilde{y}
\] (12)

where \( \tilde{y} \) is measured value, \( \hat{x} \) is estimate and \( H \) is basis function matrix. To address the least squares problem, the basis function matrix needs to be constructed as follows. Each row represents the Cartesian coordinate of knee center recorded by Kinect system and \( m \) is the number of measurements.

\[
H = \begin{bmatrix}
    x_{\text{knee},1} & y_{\text{knee},1} & z_{\text{knee},1} & 1 \\
    x_{\text{knee},2} & y_{\text{knee},2} & z_{\text{knee},2} & 1 \\
    \vdots & \vdots & \vdots & \vdots \\
    x_{\text{knee},m-1} & y_{\text{knee},m-1} & z_{\text{knee},m-1} & 1 \\
    x_{\text{knee},m} & y_{\text{knee},m} & z_{\text{knee},m} & 1 \\
\end{bmatrix}_{\text{Kinect}}
\] (13)

Next, the estimate of x-coordinate of knee center is determined using equation (12) and the estimates of y and z coordinate is calculated using the same process [84].
\[
\begin{bmatrix}
a_{1x} \\
a_{1y} \\
a_{1z} \\
a_{1c}
\end{bmatrix}_{\text{Estimate}} = (H^T H)^{-1} H^T 
\begin{bmatrix}
x_{\text{knee,1}} \\
x_{\text{knee,2}} \\
\vdots \\
x_{\text{knee,m-1}} \\
x_{\text{knee,m}}
\end{bmatrix}_{\text{Kin.}}
\begin{bmatrix}
x_{\text{knee,1}} \\
x_{\text{knee,2}} \\
\vdots \\
x_{\text{knee,m-1}} \\
x_{\text{knee,m}}
\end{bmatrix}_{\text{Vicon}}
\]

By zero model error assumption, the estimated coordinate of knee center is expressed as

\[
\hat{y} = H\hat{x}
\]

and

\[
\begin{bmatrix}
x_{\text{knee,1}} \\
x_{\text{knee,2}} \\
\vdots \\
x_{\text{knee,m-1}} \\
x_{\text{knee,m}}
\end{bmatrix}_{\text{Est.}} = \begin{bmatrix}
x_{\text{knee,1}} & y_{\text{knee,1}} & z_{\text{knee,1}} & 1 \\
x_{\text{knee,2}} & y_{\text{knee,2}} & z_{\text{knee,2}} & 1 \\
\vdots & \vdots & \vdots & \vdots \\
x_{\text{knee,m-1}} & y_{\text{knee,m-1}} & z_{\text{knee,m-1}} & 1 \\
x_{\text{knee,m}} & y_{\text{knee,m}} & z_{\text{knee,m}} & 1
\end{bmatrix}_{\text{Kin.}} \begin{bmatrix}
a_{1x} \\
a_{1y} \\
a_{1z} \\
a_{1c}
\end{bmatrix}_{\text{Est.}}
\]

### 3.2.3 Estimation Result

Plots of knee joint flexion angle and the y-coordinate of the knee center depicted in Figure 25 clearly shows the enhancement of the estimated Kinect data. The black line represents original Kinect data, the red line is Vicon data and the blue line is the estimated Kinect data. Even though this system is modeled using linear assumption between Kinect and Vicon data, the plots shows the power of least squares for model identification.

![Plots of knee joint flexion angle and the y-coordinate of the knee center](image)
As shown in Figure 26, the maximum average residual error was 22.5mm for position and 6.5 degree for knee flexion angle compared with 80.5mm and 14 degree for measured error respectively. Except for several outliers, most of average position errors are reduced with magnitudes less than 15mm [30].

Two motion capture systems – a low-fidelity/low-cost Kinect framework and the more-expensive, higher fidelity Vicon – were examined to aid quantitative knee-angle estimation in a clinically-focused squatting study.

The combination of Vicon Motion-capture with AMS/Visual-3D post-processing yielded
outstanding performance (with huge workspace, high resolution and fast sampling rate) but with limited portability and high cost. While the commercial-off-the-shelf Kinect system offers an ultra-mobile (2.2 kg) solution costing less than $100, the small workspace, relatively low resolution limits clinical or research applicability (without post-processing raw data). However, with suitable post processing, it offers potential for clinically relevant use.

### 3.3 Human Identification

#### 3.3.1 Data Collection

In human identification study, five healthy male subjects without knee injury history were studied. Subjects performed a series of 40 mild squatting trials with several practice trials given before recording. The first and last four trials among the 40 squats were recorded, and this was repeated over four days. Given performance of Kinect system highly depended on location of the subject in the workspace, the optimal environment for motion capturing was based on our previous research [30].

![Image](image.png)

**Figure 27.** (a) Data structure, (b) trajectory of hip (left/right) and knee (Left/Right) Joint
Figure 27 shows the data structure over the four days. In our study, the emphasis was solely motion of lower limb, hence ankle, knee and hip joints positions are recorded. Because the average sampling rate of Kinect system was 30 frames per second, the data was downsampled to 200 frames for each repetitive squatting cycle. One data set consisted of four trials of squatting performed by five subjects and we have total 8 sets of data and total 160 trials of data during four days’ experiment. Observed in Figure 27-(b) are knee and hip joints trajectories where it is evident the different colors corresponds to each subject exhibit significant clustering and comparable patterns.

3.3.2 Principal Component Analysis

Principle Component Analysis (PCA) is a popular tool to extract relevant information from confusing data set. PCA reduces a complex data set to a lower dimension in order to reveal the hidden, simplified structure [86].

From given data points, $x_1, x_2, \ldots, x_n \in \mathbb{R}^P$, defined the reconstruction of data in $\mathbb{R}^q$ to $\mathbb{R}^p$ and expressed as

$$f(\lambda) = \mu + v_q \lambda$$

(16)

Where $p$ is original higher dimension, $q$ is projected lower dimension, $\mu$ is mean, $v_q$ is the $p \times q$ matrix with $q$ orthogonal unit vectors and $\lambda$ is the projected low-dimensional data points.

By carefully chosen $\mu$, $v_q$ and $\lambda$, we can minimize the error between original and reconstructed (lower dimension) data as following

$$\min \sum_{i=1}^{N} \|x_n - \mu - v_q \lambda_n\|$$

(17)
Where $\mu$ is the intercept of the lower space in higher space and $\lambda_1, \ldots, n$ is the coordinate of $x$ in lower dimension ($q$), we define $\mathbb{R}^p$ plane with $\nu_q$ and $\mu$. The quantity inside the sum (Equation (17)) is the distance between the original data and the low-dimensional representation reconstruction in the original space as shown in Figure 28. The optimal intercept is the sample mean thus it is assumed as $\mu^* = 0$ and $x' = x - \mu$. Then the principle component $\nu_q$ is determined using singular value decomposition (SVD) as follows:

$$\min \sum_{i=1}^{N} \| x_n - \nu_q \nu_q^T x_n \|$$

(18)

Consider

$$X = UD\nu^T$$

(19)

Where $X$ is an $N \times p$ matrix, $U$ is $N \times p$ orthogonal matrix and the column of $U$ are linearly independent, $D$ is a positive $p \times p$ diagonal matrix with $d_{11} \geq d_{22} \geq \cdots \geq d_{pp}$, and $\nu^T$ is a $p \times p$
orthogonal matrix. From equation (19), $U$ is a low dimensional representation and $D$ reflects the variance. Lastly, $V$ is the principle component.

For low-dimensional representation reconstruction, we cut SVD at $q$ dimensions as follows:

$$X_{n \times p} = U_{N \times q} D_{q \times q} V^T_{q \times p} \quad (20)$$

### 3.3.3 Human Classification (PCA & K-NN)

![Diagram](image)

**Figure 29. Overall procedure for PCA/K-NN classification from motion capturing to validation**

Principal Component Analysis (PCA) and K-Nearest Neighbors (K-NN) method was adapted to aid in subject classification [89]. Preprocessing stages included segmentation using epochs of the squat cycle then the data was renormalized because the duration of squatting motion was not uniform across subjects and trials. For performance analysis, accuracy was obtained using a 10-fold cross-validation. Figure 29 outlines the principal stages in this process.
Principal Component Analysis (PCA) is a non-parametric statistically-based Machine Learning approach to reduce the dimension of data by restricting attention to dimensions of maximum variation to extract relevant information from confusing data sets. Since our original data lies in high dimensional space (160 x 600), substantial data is required for learning a classifier in original space. The benefits of such a statistically-based dimensionality reduction includes providing a simpler representation and revealing hidden structure of the data. Therefore the brief steps of principal component analyses is presented but the reader is referral to [86, 89, 90] for a detailed examination of the background of PCA.

In our study, we recorded and interpolated joint positions at 200 samples per squatting cycle, then we have recorded 4 days × 2 sets/subject × 4 trials/set × 5 subjects/day = 160 trials. This data was recorded into a matrix $X$, where each trial was recorded as a row of $X$. Within each row, the column data set describes Cartesian x-y-z coordinate of each of the four joints in order (left knee, left hip, right knee and right hip). This yields a 160 x 600 matrix for each joint (or a 160 x 2400 matrix if all four joints were considered) as depicted below (Equation 21-22).

$$X_{All\text{ Joints}} = \begin{bmatrix}
x_{Left\text{ Knee},1,1} & y_{Left\text{ Knee},1,1} & z_{Left\text{ Knee},1,1} & x_{Left\text{ Hip},1,1} & \cdots & z_{Right\text{ Hip},1,m} \\
 x_{Left\text{ Knee},2,1} & y_{Left\text{ Knee},2,1} & z_{Left\text{ Knee},2,1} & x_{Left\text{ Hip},2,1} & \cdots & z_{Right\text{ Hip},2,m} \\
 \vdots & \vdots & \vdots & \vdots & \ddots & \vdots \\
 x_{Left\text{ Knee},i,1} & y_{Left\text{ Knee},i,1} & z_{Left\text{ Knee},i,1} & x_{Left\text{ Hip},i,1} & \cdots & z_{Right\text{ Hip},i,m} \\
 x_{Left\text{ Knee},n,1} & y_{Left\text{ Knee},n,1} & z_{Left\text{ Knee},n,1} & x_{Left\text{ Hip},n,1} & \cdots & z_{Right\text{ Hip},n,m} 
\end{bmatrix}_{160 \times 2400}$$

$$X_{Left\text{ Hip}} = \begin{bmatrix}
x_{Left\text{ Hip},1,1} & y_{Left\text{ Hip},1,1} & z_{Left\text{ Hip},1,1} & \cdots & z_{Left\text{ Hip},1,m} \\
 x_{Left\text{ Hip},2,1} & y_{Left\text{ Hip},2,1} & z_{Left\text{ Hip},2,1} & \cdots & z_{Left\text{ Hip},2,m} \\
 \vdots & \vdots & \vdots & \ddots & \vdots \\
 x_{Left\text{ Hip},i,1} & y_{Left\text{ Hip},i,1} & z_{Left\text{ Hip},i,1} & \cdots & z_{Left\text{ Hip},i,m} \\
 x_{Left\text{ Hip},n,1} & y_{Left\text{ Hip},n,1} & z_{Left\text{ Hip},n,1} & \cdots & z_{Left\text{ Hip},n,m} 
\end{bmatrix}_{160 \times 600}$$
For analysis, the left hip-joint position data was used because it exhibited better clustering characteristics. The data-preprocessing involved first centering the data set so that the resultant had a zero mean \( X_0 = X^T - l \mu \) then the covariance matrix of the zero mean data set \( C = X_0^T X_0 \) was computed. Eigenvectors and eigenvalues were then obtained from an eigen-decomposition of the covariance matrix \( C \).

### 3.3.4 Classification Results

![Figure 30. (a) Original left hip joint trajectory, (b) Projected left hip joint trajectory of all trials](image)

To demonstrate the effect of PCA, the original data sets and data set represented in a new basis formed by the first three principal components (dimensions with maximum eigenvalues) are shown in Figure 30. Although the projected data appears well clustered than the original/raw data, it proves inadequate to allow for satisfactory classification. Hence, more numbers of principal components were employed but appropriate visualization now proves difficult. By illustrating the relationship between the variance captured and the dimension, as shown in Figure 31, fewer than 11 of 600 principal components captured 99% of the variance of the original data set.
Identifying subjects as 1 to 5, the labels were computed by K-Nearest Neighbors Method. This method uses a predetermined metric (Euclidean distance) to compute the distance between testing data and training data sets. Distances were sorted in ascending order and trials which have equivalently larger similarity return smaller distance and locate in higher order.

![Percentage of variance](image)

**Figure 31. Percentage of variance**

The number of metric (trials, K) used for comparison was decided by empirical trials, as shown as Figure 32. With the histogram of labels of the first k neighbors, the class label was decided by the most popular label [89, 90].

<table>
<thead>
<tr>
<th>Labels</th>
<th>Euclidean Distances</th>
</tr>
</thead>
<tbody>
<tr>
<td>5</td>
<td>1.0371e-15</td>
</tr>
<tr>
<td>5</td>
<td>0.2683</td>
</tr>
<tr>
<td>5</td>
<td>0.2683</td>
</tr>
<tr>
<td>5</td>
<td>0.3342</td>
</tr>
<tr>
<td>5</td>
<td>0.3753</td>
</tr>
<tr>
<td>5</td>
<td>0.3802</td>
</tr>
<tr>
<td>4</td>
<td>0.3908</td>
</tr>
<tr>
<td></td>
<td></td>
</tr>
</tbody>
</table>

![Euclidean distance and label in ascending order of ith trial and 10–fold Cross-validation](image)

**Figure 32.** (a) Euclidean distance and label in ascending order of i\(^\text{th}\) trial and (b) 10–fold Cross-validation
Our analysis utilized a 10-fold cross-validation method for accuracy calculation. The original sample was portioned into 10 equal size subsamples with each subsample used as validation data and the remaining 9 subsamples were used as training data. All observations were used for both training and validation and each observation was used for validation exactly once as described in Figure 32.

The relationship between classification rates, the number of nearest neighbors and principal components are shown in Figure 33. Higher PC and K value do not guarantee higher accuracy. The average classification rate calculated using the 10-fold cross-validation method was 95.6 % maximized when the number of nearest neighbors was 2 and number of PC was 18. The classification rate of individual subjects depends on the isolation characteristic of data cluster, which are shown in Table 2. The highest rate was 100% and lowest rate is 84.1%.

<table>
<thead>
<tr>
<th>Sub. No.</th>
<th>Rate(%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>100</td>
</tr>
<tr>
<td>2</td>
<td>100</td>
</tr>
<tr>
<td>3</td>
<td>98.9</td>
</tr>
<tr>
<td>4</td>
<td>98.9</td>
</tr>
<tr>
<td>5</td>
<td>84.1</td>
</tr>
<tr>
<td>Mean</td>
<td>95.6</td>
</tr>
</tbody>
</table>

Table 2. Classification rate of each subject (5 Subjects, four days)

![Figure 33. Classification rate with respect to varying number of first K-neighbor and order of PC (K = 2, PC = 18, Max. Rate = 95.6)](image)

With failure of classification/identification occurring at the boundary of neighboring clusters, as depicted in Figure 34, larger groups of subjects may cause lower classification accuracy as shown in Figure 34 and Table 3.
### Table 3. Classification rate of each subject (9 subjects, one day)

<table>
<thead>
<tr>
<th>Sub. No.</th>
<th>Rate(%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>100</td>
</tr>
<tr>
<td>2</td>
<td>100</td>
</tr>
<tr>
<td>3</td>
<td>100</td>
</tr>
<tr>
<td>4</td>
<td>96</td>
</tr>
<tr>
<td>5</td>
<td>100</td>
</tr>
<tr>
<td>6</td>
<td>98</td>
</tr>
<tr>
<td>7</td>
<td>80</td>
</tr>
<tr>
<td>8</td>
<td>98</td>
</tr>
<tr>
<td>9</td>
<td>64</td>
</tr>
<tr>
<td>Mean</td>
<td>92.9</td>
</tr>
</tbody>
</table>

**The classification rate for extended data (9 subject, one day) was reduced to 92.9% compared with 95.6% (5 subjects, four days). The results suggest the short time-scale noisiness of the Kinect sensor creates challenges for real-time applications (as evident by the rapid fluctuations of the skeletal pose estimates).**

However, our operant hypothesis was that the geometric consistency of the human musculo-skeletal system and statistical averaging may allow exploitation of the Kinect data to realize considerable accuracy and consistent performance. This was examined in detail in our work to determine subject-geometric-parameters and reconfirmed by the subject classification rate. Classifications rates approached 95.6% on average for data recorded over four separate days.

Figure 35 shows classification rates of various data (trajectory of hip, knee, hip & knee joint). By using hip joint data, we obtained higher classification rates than knee joint and when recruiting diverse data (combination of knee and hip joint trajectory), it improved classification rates 96.4% was observed [31].
The low-cost Commercial-off-the-shelf Kinect System system has many benefits over high-end motion capture systems, including markerless operation, portability, simplified PC-interfaces and most importantly low-cost. While the small work-volume and low spatio-temporal resolution created challenges, the Kinect offers significant benefits for deployment of home-based internet-monitored progressive exercise regimen.

3.4 Static Measurement by Wii Balance Board

Newton-Euler formulation treats each link in turn and derives the equations describing its linear and angular motion. Because each link is coupled to other links, these equations describe coupling forces and torques. By doing forward-backward recursion, we can determine all coupling terms and a description of the manipulator as a whole [82]. Refer to 3.1.3 for lower limb kinematics and dynamics, and section 2.3 for Wii Balance Board performance.
For consistent measurement, the subject’s foot was fixed to the Wii balance board using foot stopsers as shown in Figure 36 – (b). Because of Wii Balance Board limitation, only the normal force (Fy) and mediolateral (Mx) and anteroposterior (Mz) moments were measured. As a result, the magnitude of displacement and load at the knee center in the other directions were relatively small and with the variance within the measurement of the system. Wii balance board communicated with host PC via Bluetooth using a custom Matlab GUI developed in house. Knee force and moments were displayed in real time and the data was record it for further analysis as shown in Figure 36 – (a).

![Knee Joint Angle](image1)

![Force/Moment at Knee Center](image2)

(a) (b)

Figure 36. (a) Measurement result and (b) Wii balance board setup for motion capture

Because of the slow movement task, the relationship between time and inertial mass were considered negligible so we assumed a quasi-static condition in our analysis.
3.5 Knee Performance Measurement

Optical tracking system and force platform introduced in the background section are state-of-the-art technology for capturing human motion, but it has a limit on measuring dynamic responses of specific joints such as knee and ankle. Roy et al. [91] developed ankle robot to estimate human ankle stiffness which is intended for rehabilitation of the human ankle.

For an in-depth study for knee function, we developed the Knee Performance Measuring Device which consisted of hexapod, 6-DOF force transducer (ATI, Delta) and optical tracking cameras (OptiTrack, Flex3) as shown in Figure 37.

![Figure 37. Knee Performance Measuring Device setup with saw bone model (a) and (b) human subject](image)

The hexapod provides independent and active assistance 6-DOF motion to the knee joint while the ankle joint was fixed to knee support. A multi-axis force transducer (ATI Delta [92]) fixed in between the hexapod and knee support measured all six components of force and torque rigidly. By tracking reflective markers attached on anatomical landmarks of lower limb, the optical tracking system captures tibiofemoral joint motion. To verify the measurement system,
we tested it with standard specimen (Aluminum 6061, 520mm, E = 69GPa, maximum deflection 10mm) which is inspired from Material Testing System (MTS) as shown in Figure 38.

![Figure 38. Setup for specimen test and Force vs. deflection plot (a) without and (b) with knee support](image)

Theoretical normal force at the end of the beam was 5 N and the measured force was 4.2N for a 10mm deflection. We suppose that most of error caused from tolerance of assembled measurement and support system. To test the effect of knee support on measurement result, we performed same test both with and without knee support and the difference of force at maximum deflection is 0.2N.
For consistency, the shank and thigh were aligned as shown in Figure 39. The knee center was aligned with the hip center and the ankle center was aligned with the knee center, making the axes of the femur and tibia parallel and perpendicular to the ground respectively. After measuring maximum knee extension angle, the hexapod moved in three directions and the reaction force and torque were recorded using the 6-DOF force transducer.

Figure 39. Measurement process for Knee Performance Measuring Device

Figure 40. GUI for knee performance measuring device
All processes were controlled and monitored manually or automatically using the custom Matlab GUI, as shown in Figure 40.

Figure 41. (a) Position & force vs. time and (b) force vs. deflection

Preliminary tests were undertaken with healthy subjects and both kinematic and static data, were recorded in real time (1000Hz) as shown in Figure 41. Figure 41 – (a) depicts the relationship between absolute spatial position of the knee and concomitant forces with the slope of the plots in Figure 41 – (b) representing knee stiffness in all three directions.
Neuromuscular response to functional knee bracing was examined by Ramsey et al. [93] to investigate the effect of a commercial functional knee brace during activity of anterior cruciate ligament deficient subjects. For the purpose of feasibility, evaluation of our Knee Performance Measuring Device, we measured reaction forces both with and without knee strap as shown in Figure 42 and one can clearly see the effect of knee strap.
4 Smart Brace Design- Kinetostatic Design Optimization

The majority of current portable orthotic devices and rehabilitative braces fall under the ‘passive’ class — focused on providing stability, facilitating constrained motions, or helping maintain alignment of the joints. The convergence of miniaturization of sensing/actuation with advances in computing and communication has facilitated creation of a class of smart, compact and lightweight and wearable rehabilitation devices, that are highly adaptive, versatile and reprogrammable in nature. This has spawned development of classes of ‘semi-active’ (with adjustable but passive springs/dampers) and even ‘active’ (powered actuator) rehabilitation braces seeking to augment muscle function and promote faster recoveries. Most of the early emphasis was placed on the mechatronic integration and packaging aspects. Currently a wide variety of designs exist for embodiment (physical mechatronic implementation) of such smart adjustable rehabilitation braces/exoskeletons to address rehabilitation-related requirements including safe and dependable physical interaction, true wearability and portability, and user aspects such as acceptance and usability.

From this perspective, we developed a low-cost real-time articulated electromechanical system capable of controlled bidirectional kinesthetic (motion and force) interactions in patients. Implementation of a low-cost prototype will allows us to develop and evaluate clinically-relevant yet quantitative performance assessments (with adequate specificity, selectivity, resolution, repeatability) of the degree of functional recovery.
4.1 Knee Brace

Exoskeletons/braces, which are designed to augment human performance are comprised of an articulated mechanical structure, with actuators, or visco-elastic components, sensors and control elements. Compared to exoskeletons, braces employ simpler and less complex designs and focus primarily on passive structural equilibration of the externally applied loads.

Knee braces can be used to partially compensate for disrupted bony or soft tissue elements. They can also be used postoperatively or prophylactically to protect these elements during healing or regular activities, respectively. The ideal knee brace should be designed under the principles such as not interfering with normal knee function, not increasing the risk of injury to the other parts of the lower extremity in addition to producing synergetic effect with physiological components.

The types of knee braces are categorized into five brace categories: prophylactic, rehabilitative, functional, off-loader, and patellofemoral knee braces as shown in Figure 43. Prophylactic knee braces (PKB), single or bilateral uprights which have a single, double or polycentric hinge are common design. These braces protect MCL and ACL against direct lateral blows to the knee by dispersing the impact force along the lines of the lower extremity. The usual rehabilitation braces are used postoperatively to restrict range of motion whereas functional braces are prescribed to limit abnormal knee rotational and translational forces. The off-loader braces apply an external valgus-bending moment to the knee that in theory unloads the medial compartment and mediates pain relief for OA patient. The patellofemoral knee brace mediates improved or correct patellar tracking and alters quadriceps function [8].
4.2 Classification of Exoskeleton System

Herr and Dollar & Herr [94, 95] offer excellent state-of-the-art surveys of lower extremity exoskeletons and active orthoses. Architecturally, exoskeletons and braces may be differentiated into serial- and parallel- chain systems. Serial-limb exoskeletons, introduced in Figure 44, showed improved performance in selected criteria (such as jumping height). However, their broad applicability for augmentation, energy efficiency or running speed has not been effectively studied.
Parallel exoskeleton designs focus on load bearing and effective transferring of large loads (including body weight) through the exoskeleton to the ground. For example, the elastic exoskeleton developed at MIT Biomechatronics Group [98] is capable of transferring body weight directly to the ground during stance, and therefore lowers metabolic demand. Other examples of advanced parallel-limb exoskeletons, that augment load-carrying capacity are also illustrated in Figure 45.

![Figure 45](image)

Figure 45. Examples of Parallel-limb Exoskeleton to augment load-carrying capacity, (a) Hopping Exoskeleton [98], (b) BLEEX System by Berkeley Robotics & Human Engineering Lab [47] and (c) HULC System by Lockheed Martin [48].

Parallel-limb exoskeletons that augments joint torque/force augmentation is shown in Figure 46. Such exoskeletons are targeted for performance-augmenting, rehabilitation and assistance for deficit gait. Comparing with the load-carrying exoskeleton, this type of exoskeleton/brace does not transfer load to the ground but focuses more on capability to reduce joint pain and increase joint strength.

Although remarkable progress has been made in developing powered robotic exoskeletons and powered orthoses, many design and implementation challenges still plague deployments.
Most powered exoskeleton devices tend to be heavy, bulky, noisy and can produce limited torque and power. Another limit of design is lack of consideration in effective mechanical interface which can cause discomfort to user and decrease efficiency of system [95].

![Figure 46. Examples of Parallel-limb Exoskeleton to augment joint force/torque (a) HAL System by Cyberdyne [99], (b) Quasi-Passive Knee Exoskeleton [46, 100] and (c) Active Ankle-foot Orthosis [100].](image)

Unfortunately, effective function of the knee brace/exoskeleton depends both on the designs as well as proper fitting to the patient. If the exoskeleton is not optimally matched to the patient’s limb, distal brace/exoskeleton migration and poor torsional stability may result. For the ideal knee exoskeleton/brace design, individual mobility, alignment, clearance, stability, stiffness, simplicity, adjustability, energy efficiency, weight and durability are all critical criteria. Although a flexible structure offers better adjustability and alignment, only the rigid architecture provides sufficient stiffness to prevent knee buckling. Weight, durability and energy efficiency are closely related with simplicity of design. The excessive weight of the exoskeleton affects gait pattern, applies undesirable load at the ankle/hip joint, and causes high energy consumption for the user.
4.3 Literature Review

Quintero et al. introduced a powered lower-limb orthosis which augmented hip and knee moments to persons with spinal injury to assist with walking [101]. Dollar and Herr described a knee-brace in which a motorized mechanism actively introduced and removed a spring in parallel with the knee joint [46]. BLEEX exoskeleton system utilized 7 d.o.f per leg, four of which were powered by linear hydraulic actuators [47]. Sankai et al. proposed a calibration method to correlate EMG to joint torque in order to follow operator intent [99]. The electro-rheological fluid (ERF) smart brake facilitates highly tunable resistive torque capabilities through a variable damper component [102]. A biomimetic active agonist-antagonist structure designed to reproduce both positive and negative work phases of the natural joint was developed by Martinez-Villalpando et al. [46]. Össur introduced the first bionic leg that combines a powered ankle and an adaptive microprocessor-controlled knee joint [45]. Lockheed Martin developed the HULC an untethered, hydraulic-powered anthropomorphic exoskeleton with the ability to carry 200 lb. loads [48]. Semi-active powered exoskeletons are also being developed, e.g. Ward et al. [103] created an adjustable spring actuated ankle orthosis to control foot drop during swing.

In recent years, researchers have examined the development of a systematic methodology to improve both brace- and fixation-design [104-106]. However, while systematic, such approaches remain qualitative at best. Schiele et al. [106] examined the kinematic design of an upper limb exoskeleton which was comfortable to wear and did not create residual force and torque although misalignment exists. Instead of imitating the kinematic structure of a human limb, an alternative moving system bridging in parallel with the human joints was introduced.

Kinematically similar chains that are not perfectly identical may generate uncontrollable and undesirable motion and force although the degree of hyperstaticity is reduced by soft skin and
elastic straps. Jarrasse et al. [104, 105] presented a methodology for designing the fixation for exoskeleton to the human limb. They proposed a set of design rules for the hyperstaticity problem when human limb and exoskeleton are connected with multiple fixation points. Although design rule for exoskeleton was introduced, it is limited to the choice of degree-of-freedom for actuation and fixation parts. Once again, details of configuration of linkage (such as type and order of joints) were chosen based on intuition, iteration and practical experience of designers.

4.4 Kinetostatic Design Optimization

There are many choices of various mechanism configuration and mechanism design parameters that need to be made in order to design a multi-body system. The general design problem will be solved by dimensional synthesis of mechanisms, the process of determination of the dimensions of mechanism to match desired specifications which offer a systematic means for selection of unknown parameters.

Given a set of task specifications and the type of mechanism, an optimization problem relating the parameters of the device to the set of desired specifications can be formulated and solved. Specifically, we match the desired specifications on end-effector positions and forces. Other types of design specifications can now also be explored. However, the resulting solution mechanism satisfies all these desired specifications only in the least squares sense without guaranteeing exact satisfaction of any specification.

Greater structure is added to this problem by employing Precision Point Synthesis (PPS). The requirement to match specifications exactly at precision points creates constraints between the various parameters of the mechanism which aids the final selection of the parameters of the designed device. The design constraint equations, themselves, are created from the equations of
loop-closure for end-effector position specifications or by application of the principle of virtual work for end-effector force specifications.

The requirement for exact matching of more specifications at precision points creates more constraints. A limit on the number of specifications/points is finally reached and a simultaneous solution of the constraint system yields a unique mechanism. While exact matching at the greatest number of precision positions is desirable, the gains in terms of precise specification matching at many points may be offset by the disadvantage of having to solve a set of nonlinear equations. Hence, in what follows, we consider: the specification of less than maximum feasible number of precision points; a suitable partitioning of all the variables into dependent and independent variables; a suitable specification of the independent variables; leading up to the formation of a linear system of equations in terms of the dependent variables.

Desired mechanism specifications are typically presented in the form of a discrete set (of positions, orientations and forces) to be achieved or approximated as closely as possible by the manipulatory device. In optimization based schemes, it is required to match the design specifications over the range of travel of the mechanism. Hence, some method of interpolating in between the discrete specifications is required to obtain a continuous “desired curve”. The “generated curve” (or alternatively “mechanism curve”) refers to the curve generated by the mechanism whose parameters were determined earlier with the aid of the precision point equations. In general, there is no correspondence between the generated curve and the desired curve except at the precision points due to the different parameterizations. We consider different approaches for the identification of two sets of points that have a one-to-one correspondence to each other based on certain criteria. A point based measure of discrepancy is now required to enable the definition of a cumulative measure of discrepancy over the finite number of
correspondence points which now forms the objective function for the optimization. In the case of uniform specification matching, a summed squares of the discrepancy between the correspondence points is used. For mixed design specification matching requirements, a scale dependent left invariant metric reflecting engineering considerations are used.

The specification of values for the independent variables coupled with the solution of the linear system of equations for the dependent variables creates candidate mechanisms. An optimization scheme over the different candidate mechanisms yields the parameters of the optimal mechanism which satisfies the design specifications exactly at the selected precision points and in the least-squares sense elsewhere. The principal advantage of this approach is the ability to add structure to the problem while offering adequate flexibility/choices to the designer.

![Kineto-static Optimization Process](image)

**Figure 47. Kineto-static Optimization Process**

The key steps to this process are: creation of precision point synthesis design constraints and solution to obtain feasible mechanism candidates; a method of interpolation between desired specifications; a method of obtaining correspondence between the desired specifications and
obtained performance; and a measure of error to minimize between the two are summarized in Figure 47 [32, 107, 108].

4.4.1 Kinetostatic Design Optimization of Knee Brace

Matching the geometric-, kinematic-, and dynamic-performance of the brace to the individual is of prime importance [93, 109, 110]. Most research employ a top-down approach to assess and/or design knee braces by customizing state-of-art commercial knee-brace and selecting design parameters to improve performance. Most often, in practice, knee braces are designed and prescribed on the basis of an expert’s knowledge and subjective experience.

Significant nonlinearities are evident in knee-kinematics and kinetics. A prime example is that knee flexion/extension occurs about an instantaneous center-of-rotation which follows a spatial time-varying trajectory (rather than a spatially-fixed center-of-rotation). This observation has in the past led up to creation of various polycentric hinge designs (based off innovative 6-bar designs [111]). Yet to the best of our knowledge, systematic and quantitative efforts at knee brace design/prescription to match both desired motion- and force- interactions have lagged behind. In our work, we showcase some of our efforts at furthering a kinetostatic-design approach to allow for better matching of motion- and force-interactions.

The design problem is formulated on the framework of dimensional synthesis of mechanisms [56, 112]. In particular, given task specifications and the type of mechanism, an optimization problem can be formulated to determine the set of parameters to match desired specifications. This offers a systematic process for selecting large sets of unknown parameters. However, the resulting solutions satisfy all these desired specifications only in the least-squares sense without guaranteeing exact satisfaction of any specification.
Figure 48. The process of Kinetostatic optimization for articulated(four-bar linkage) knee brace

Greater control is added when employing Precision Point Synthesis (PPS). Matching specifications exactly at precision points create constraints between the various design parameters, which helps in the final selection of design parameters. These constraints can be combined with previously described optimization-based frameworks to create a constrained design optimization problem for mechanism synthesis. Typically, the constraints are derived solely from kinematic considerations. These are typically integrated back into the optimization problem using a penalty formulation.

In general, the above problem entails a simultaneous determination of optimal kinematic and static parameters. However, for discussion in the next sections, we assume the optimal mechanism parameters have been computed previously by a kinematic path-following optimization scheme. This enables us to focus solely on the determination of the spring parameters to satisfy the desired static constraints.
In contrast, this study intended to employ additional constraints relating the joint actuation and end effector forces by applying the principle of virtual work to the articulated subsystem [56, 107]. Further, we utilized a constrained optimization solution that emphasizes: (i) the partial specification of design requirements on the end effector; (ii) a suitable partitioning of all the variables into dependent and independent variables; and (iii) explicit creation/solution of a linear system of equations in terms of the independent variables. Optimization over the independent variables yields different candidate mechanisms, which satisfies the design specifications exactly at selected precision points and in the least-squares sense elsewhere. The principal advantage is the ability to add structure to the problem while offering adequate flexibility/choices to the designer. Finally, we note that while the current emphasis is on matching the desired end-effector motion/force specification and many other types of design specifications can also be explored.

4.4.2 Kinematic Trajectory Objective

The tibio-femoral joint motion was tracked using an Vicon motion capture system, with kinematic data post-processed using Anybody Modeling system (AMS) which is based on kinematic global optimization of human model [113]. Spatial tibiofemoral motion, as depicted in Figure 49, and the relative motion between shank (tibia) and thigh (femur) was used for the desired trajectory of knee brace. As shown in Figure 49, the relative motion of femur and tibia is spatial, so the trajectory of lateral and medial uprights should be different relative to each other. In our study, we examined designs that permitted the lateral and medial uprights to follow separate trajectories, which conventional symmetric knee braces are not capable of achieving.
4.4.3 Architecture Selection & Design Framework

A four-bar architecture with torsional springs at each joint was chosen as the basic articulated knee brace design. The knee brace consists of lateral and medial uprights and each upright had independent four-bar linkage, which allowed for variation in the kinematic, static and configuration parameters for each. The ground linkage was attached to the thigh support. The coupler link was assumed to be pinned to the shank at an appropriate coupler point (as shown in Figure 50).

Suitable selection of the various mechanism parameters was critical, which included both kinematic parameters such as link-lengths and initial configuration, as well as static parameters such as spring constants and their preloads. The overall problem was considered a merger of the two interlinked stages. The kinematic stage entailed the selection of the parameters to permit the end-effector follow a desired pre-specified path relative to the fixed link (which served as the base). However, the force interaction between the linkage and its environment also became critical to performance. In this case, the goal of the articulation was to guide the attached knee brace through several positions while supporting a set of specified external loads. Hence, in the
static stage, we examined the selection of the optimal parameters to support these external loads to the largest extent by structural equilibration, building on our earlier work [56, 107].

4.4.4 Kinematic Synthesis

We pursued a 2-stage (sequential) kinetostatic synthesis and design optimization process. The first stage was a kinematic-design optimization with a 2-precision-point synthesis for the four-bar knee brace. Figure 51 depicts the various parameters of the four-bar linkage: \( Z_2(\theta_2, z_2) \), \( Z_4(\theta_4, z_4) \), \( Z_5(\theta_5, z_5) \), \( Z_6(\theta_6, z_6) \), \( \alpha_2 \), \( \alpha_4 \), \( \beta \), \( \phi \), \( M \), \( N \). \( R_1 \) and \( R_2 \) are determined by problem description and we specified the base joint vectors \( M \) and \( N \) as free choices and then solved kinematic loop closure equations for the other unknowns.

The loop-closure equations of a fourbar mechanism are:

\[
\begin{bmatrix}
Z_2 \\
Z_6 \\
Z_4 \\
Z_5
\end{bmatrix} =
\begin{bmatrix}
1 & 1 & 0 & 0 \\
e^{i\alpha_2} & e^{i\beta} & 0 & 0 \\
0 & 0 & 1 & 1 \\
0 & 0 & e^{i\alpha_4} & e^{i\beta}
\end{bmatrix}^{-1}
\begin{bmatrix}
R_1 - M \\
R_2 - M \\
R_1 - N \\
R_2 - N
\end{bmatrix}
\]

(23)
To minimize the interference of natural knee motion into knee brace, the trajectory of knee brace motion must be coincident with desired trajectory by optimizing kinematic parameters, $\alpha_2, \alpha_4, \beta$. The objective function is taken to be the structural error between the desired and actual path computed using arc length based correspondence points. The design variables are $\alpha_2, \alpha_4, \beta$ and optimization objective function is expressed as:

$$
\text{Min}_{\alpha_2, \alpha_4, \beta} \sum_{i=1}^{N} \left( (P_{x_i} - q_{x_i})^2 + (P_{y_i} - q_{y_i})^2 \right)
$$

where $P$ is point on desired curve and $q$ is point on the actual path of end-effector. In Figure 52, desired trajectory from AMS kinematic analysis and optimized configuration of fourbar are shown. The end-effector of the linkage pass two precision points marked as red circle and follows the desired curve with minimal deviation.

Figure 51. Two precision point synthesis diagram
4.4.5 Static Synthesis

This static optimization stage is followed by a kinematic design optimization which includes a static precision point synthesis stage. The static-precision equations are obtained by application of the principle of virtual work.

By the principle of virtual work, a system is to be in equilibrium and the total virtual work done by external forces during a virtual displacement must be zero. The external forces associated with the fourbar linkage are shown in the Figure 53. Let $T_{Spring,i}$ be the spring torques applied at joints required to equilibrate the external loadings where $F_x$, $F_y$, $M_z$ are applied force and the moment at the end-effector $P$ and torsional springs are located at the four joints $O_2, O_3, O_4$ and $O_5$. By equating the sum of virtual work done by the individual forces and torque to zero, we get,

$$T_{Reaction,2} \delta \alpha_2 + T_{Spring,2} \delta \sigma_2 + T_{Spring,3} \delta \sigma_3 + T_{Spring,4} \delta \sigma_4 + T_{Spring,5} \delta \sigma_5 + F_x \delta P_x + F_y \delta P_y + M_z \delta \phi = 0$$

(25)
By differentiating the forward kinematic equation of fourbar mechanism, we obtain:

\[
\begin{bmatrix}
\delta \theta_2 \\
\delta \theta_3 \\
\delta \theta_4
\end{bmatrix} = \frac{1}{z_2 \sin (\theta_4 - \theta_2)} \begin{bmatrix}
z_5 \sin (\theta_3 - \theta_4) \\
z_2 \sin (\theta_3 - \theta_2) \\
z_4 \sin (\theta_3 - \theta_4)
\end{bmatrix} \delta \theta_2 = \begin{bmatrix}
\delta \theta_2 \\
A \cdot \delta \theta_2 \\
B \cdot \delta \theta_2
\end{bmatrix} = \begin{bmatrix}
\delta \alpha_2 \\
A \cdot \delta \alpha_2 \\
B \cdot \delta \alpha_2
\end{bmatrix},
\]

(26)

Let 
\[
A = \frac{z_2 \sin (\theta_4 - \theta_2)}{z_3 \sin (\theta_3 - \theta_4)}, \quad B = \frac{z_2 \sin (\theta_3 - \theta_2)}{z_4 \sin (\theta_3 - \theta_4)}
\]

The displacement, \(\delta P\) may be written in terms of \(\delta \alpha_2\) as:

\[
\delta P = \delta \alpha_2 (-z_2 \sin \theta_2 \delta \theta_2 - z_6 \sin (\theta_3 + \gamma_6) A) i + \delta \alpha_2 (z_2 \cos \theta_2 \delta \theta_2 + z_6 \cos (\theta_3 + \gamma_6) A) j
\]

(27)

and the angular extension of each spring \(\sigma_i\) is:

\[
\sigma_2 = \theta_{2,initial} + \alpha_2 - \Omega_2
\]

\[
\sigma_3 = (\theta_{3,initial} + \beta - \Omega_3) - (\theta_{2,initial} + \alpha_2 - \Omega_2)
\]

\[
\sigma_4 = \theta_{4,initial} + \alpha_4 - \Omega_4
\]
\[ \sigma_5 = (\theta_{5, \text{initial}} + \beta - \Omega_5) - (\theta_{4, \text{initial}} + \alpha_4 - \Omega_4) \]

where \( \Omega_i \) are preloaded angle of torsional springs. The virtual angular displacements \( \sigma_2, \sigma_3, \sigma_4 \) and \( \sigma_5 \) can be expressed in terms of independent virtual angular displacement \( \alpha_2, \alpha_4, \beta \) as

\[ \frac{\partial \sigma_2}{\partial \alpha_2} = 1, \quad \frac{\partial \sigma_3}{\partial \alpha_2} = A - 1, \quad \frac{\partial \sigma_4}{\partial \alpha_2} = B, \quad \frac{\partial \sigma_5}{\partial \alpha_2} = A - B \]

Where \( A = \frac{z_2 \sin(\theta_4 - \theta_2)}{z_3 \sin(\theta_2 - \theta_4)}, B = \frac{z_2 \sin(\theta_3 - \theta_2)}{z_4 \sin(\theta_3 - \theta_4)} \)

Thus equilibrium equation of virtual work can be rewritten as

\[ \left( T_{\text{reaction}} + F_x \frac{\delta P_x}{\partial \alpha_2} + F_y \frac{\delta P_y}{\partial \alpha_2} + M \frac{\delta \phi}{\partial \alpha_2} - k_2 \sigma_2 \frac{\partial \sigma_2}{\partial \alpha_2} - k_3 \sigma_3 \frac{\partial \sigma_3}{\partial \alpha_2} - k_4 \sigma_4 \frac{\partial \sigma_4}{\partial \alpha_2} - k_5 \sigma_5 \frac{\partial \sigma_5}{\partial \alpha_2} \right) \frac{\partial \alpha_2}{\partial \alpha_2} = 0 \tag{28} \]

In passive knee brace, the system is static equilibrium at every configuration and external force is balanced force of gravitational, external, muscle and ligament forces at knee so reaction torque, \( T_{\text{reaction}} \), at driving joint can be considered as zero. By considering fixation convenience in practice, let us assume that there is a revolute joint and no friction between hinged upright and upper limb segments, shank and thigh so it results in zero moment at the end effector of fourbar linkage. Hence we obtain:

\[ F_x \frac{\delta P_x}{\partial \alpha_2} + F_y \frac{\delta P_y}{\partial \alpha_2} = k_2 \sigma_2 \frac{\partial \sigma_2}{\partial \alpha_2} + k_3 \sigma_3 \frac{\partial \sigma_3}{\partial \alpha_2} + k_4 \sigma_4 \frac{\partial \sigma_4}{\partial \alpha_2} + k_5 \sigma_5 \frac{\partial \sigma_5}{\partial \alpha_2} \tag{29} \]

The spring constants are free-chosen variables and balanced force \( F \) is a nonlinear function of \( \alpha_2 \) and linear function of spring preload, \( \Omega_i \). From Equation 26:

\[ F_x \cdot e + F_y \cdot f = k_2 \cdot a(\alpha_2, \Omega_2) + k_3 \cdot b(\alpha_2, \Omega_3) + k_4 \cdot c(\alpha_2, \Omega_4) + k_5 \cdot d(\alpha_2, \Omega_5) \tag{30} \]
For a four static-precision position problem, the spring constants and spring preloads are calculated by substituting precision forces – the specified forces \((F_i(\theta_2))\) at the precision points, \(\alpha_2\). The desired spring constants \(k_i\) and corresponding spring preload angles \(\Omega_i\) can be decided by solving:

\[
\begin{bmatrix}
  k_2 \\
  k_3 \\
  k_4 \\
  k_5 \\
\end{bmatrix} = 
\begin{bmatrix}
  a_1(\alpha_2, \Omega_2) & b_1(\alpha_2, \Omega_3) & c_1(\alpha_2, \Omega_4) & d_1(\alpha_2, \Omega_5) \\
  a_2(\alpha_2, \Omega_2) & b_2(\alpha_2, \Omega_3) & c_2(\alpha_2, \Omega_4) & d_2(\alpha_2, \Omega_5) \\
  a_3(\alpha_2, \Omega_2) & b_3(\alpha_2, \Omega_3) & c_3(\alpha_2, \Omega_4) & d_3(\alpha_2, \Omega_5) \\
  a_4(\alpha_2, \Omega_2) & b_4(\alpha_2, \Omega_3) & c_4(\alpha_2, \Omega_4) & d_4(\alpha_2, \Omega_5) \\
\end{bmatrix}^{-1} 
\begin{bmatrix}
  F_{x,1} \cdot e_1 + F_{y,1} \cdot f_1 \\
  F_{x,2} \cdot e_2 + F_{y,2} \cdot f_2 \\
  F_{x,3} \cdot e_3 + F_{y,3} \cdot f_3 \\
  F_{x,4} \cdot e_4 + F_{y,4} \cdot f_4 \\
\end{bmatrix} \tag{31}
\]

Thus Equation 31 gives us freedom for selection of spring-preloads.

### 4.4.6 Optimization Results

The static optimization problem statement for fourbar mechanism with four torsional springs may be written as:

\[
\begin{aligned}
M in & \sum_{i=1}^{N} \left( (F_{xi} - g_{xi})^2 + (F_{yi} - g_{yi})^2 \right) \\
\end{aligned} \tag{32}
\]

where \(F\) is desired force and \(g\) is actual force. The desired force profiles of medial and lateral uprights are chosen by precision forces selection which minimizes balanced forces and moments at knee. As shown in Figure 54, actual fourbar force curve passes through precision force specifications and seeks to minimize the discrepancy between desired and actual curves.
As a result of static synthesis and optimization, the minimum spring constants and spring preload within specific range are decided and results are depicted in Figure 54. In knee brace design, kinematic objective has priority over the static objective and hence, we adopted a sequential kinetostatic optimization procedure. In a more general case, it is possible to treat a weighted sum of kinematic- and static-objectives within a combined kinetostatic optimization objective function as shown below.

\[
\text{Min} \left( \sum_{i=1}^{N} w_1 \left( (P_{xi} - q_{xi})^2 + (P_{yi} - q_{iy})^2 \right) + w_2 \left( (F_{xi} - g_{xi})^2 + (F_{yi} - g_{iy})^2 \right) \right) \quad (33)
\]

Thicker lines in Figure 55 are balanced forces and moments, which are summation of load at knee and force/moment applied by designed knee brace. The maximum absolute values of those are lower than loads without knee brace as desired from our optimization efforts.
The desired force profile depicted in Figure 55 can be chosen for specific clinical purposes (such as assigning higher weight for antero-posterior force or medio-lateral moment). The desired force/moment profile can be designed and optimized to customize balanced knee loads by using weighted objective function as given below.

\[
F = w_1 \cdot \max(F_{\text{Anteroposterior}}^2) + w_2 \cdot \max(F_{\text{Mediolateral}}^2) + w_3 \cdot \max(F_{\text{Proximodistal}}^2) + w_4 \cdot \max(M_{\text{Lateral}}^2) + w_5 \cdot \max(M_{\text{Axial}}^2) + w_6 \cdot \max(M_{\text{Flexion}}^2) \tag{34}
\]

The result of the weighted optimization for one subject is shown in Figure 56. The plots indicate the decreasing medio-lateral moment (due to higher weights) while compromising other moments.
Figure 56. Balance moments at knee for higher weightage on medio-lateral moments

Thinner dashed red lines are pure reaction medio-lateral moment and thicker dashed red lines are balanced moment for which higher weightage was considered. It results in reducing maximum medio-lateral moment from 10N to 5N. Balanced moments presented in Figure 55 for even assigned weightage show less capability to reduce medio-lateral moment compared with the case in Figure 56 [32].

The systematic design of articulated knee brace offers many challenges which we address in our work. One contribution of our work is to use human motion/force data for customizing articulated knee brace design. Both kinematic and static measurements (optical marker trajectories and reaction forces) are used to decide the parameters of knee brace including link lengths, configurations, spring constants and preloads. Secondly, most commercial and research purpose knee braces focus on symmetric motion of lateral and medial upright (hinge). In contrast, our approach/design is able to generate asymmetric motion with two independently articulated (four-bar) linkages to better match the real tibio-femoral motion. Finally, brace-design process today depends only on expert’s experience, knowledge and intuition that are not easily
quantifiable. The quantitative knee-bracing design, espoused in this work, seeks to address this limitation by customizing kinematic- and static- parameters of the knee brace based on human motion/force data.
5 Smart Brace Design - Screw Theoretic Design Optimization

In this section, we simultaneously introduce a systematic process by using a twist- and wrench-based modeling/analysis to evaluate various alternate exoskeleton designs. This process is examined in the context of a case study for developing optimal configuration design, and fixation of a knee exoskeleton. The results are evaluated using a prototype 3D printed exoskeleton for use with a saw–bones knee joint model – which is then physically tested in a scaled knee brace test.

5.1 Screw theory

Rigid body motion can be moved one position to the other by a movement consisting of rotation about an axis followed by translation parallel to the axis and this motion is called as screw motion as shown in Figure 57. The infinitesimal version of a screw motion is called a twist which provides instantaneous linear and angular velocity of a rigid body. Any system of force acting on a rigid body can be replaced by a single force applied along an axis, combined with a torque about the same axis. Such a force is referred to as a wrench which is dual to twist [114].

![Figure 57. General screw motion](image-url)
The relative configuration of a moving frame \( \{E\} \) relative to a fixed frame \( \{F\} \) is defined by the homogeneous transformation expressed in a 4x4 matrix form as

\[
g_{FE}^F = \begin{bmatrix}
R_E^F & \bar{d}_E^F \\
0 & 1
\end{bmatrix}
\]  

(35)

where \( R_E^F \in SO(3) = \{ R \in \mathbb{R}^{3\times3} : R R^T = I, \det(R) = +1 \} \) is a rotational matrix and \( \bar{d}_E^F \in \mathbb{R}^3 \) is a displacement vector. Thus, the configuration of a rigid body in a three-dimensional Euclidian task space can be represented as

\[ SE(3) = \{ A : A = \begin{bmatrix} R & \bar{d} \\ 0 & 1 \end{bmatrix}, \bar{d} \in \mathbb{R}^3, R \in SO(3) \} \]

The velocity of a rigid body with respect to the body frame can be specified by

\[
\mathbf{V}_{FE}^E = g_{FE}^1 \dot{g}_{FE}
\]

(36)

The manipulator Jacobian can also be used to describe the relationship between wrenches applied at the end-effector and joint torques. The net work performed by applying a body wrench \( F_t \) over an interval of time \([t_1, t_2]\) is

\[
W = \int_{t_1}^{t_2} \mathbf{V}_{FE}^E \cdot F_t dt
\]

where \( \mathbf{V}_{FE}^E \) is the body velocity of the end-effector. The work will be the same as that performed by the joints, and hence

\[
W = \int_{t_1}^{t_2} \dot{\theta} \cdot \tau dt = \int_{t_1}^{t_2} \mathbf{V}_{FE}^E \cdot F_t dt
\]

(37)

Since this relationship must hold for any choice of time interval, the integrands must be equal. Using the manipulator Jacobian to relate \( V_{FE}^E \) to \( \dot{\theta} \), we have

\[
\dot{\theta}^T \tau = \dot{\theta}^T (J_{FE}^E)^T F_t
\]

(38)

It follows that since \( \dot{\theta} \) is free...
\[ \tau = (J^E_{FE})^T F_t \] 

If a wrench causes no joint torques, it must be resisted by structural forces generated by the mechanism. Such a situation occurs when \( F \) lies in the null space of \((J^E_{FE})^T\). In this case, the force balance equation is satisfied with \( \tau = 0 \); the resisting forces are supplied completely by the robot’s mechanical structure [114, 115].

### 5.2 Design Candidates

We assume that an exoskeleton consists of serial chain of links connected by rotary (revolute) or linear (prismatic) joints. This serial exoskeleton is assumed to be fixed in parallel to the articulated multi-body skeleton of the patient limb. The selection of the links, their dimensions and attachment via joints and ultimately actuation determines the efficacy any articulated mechanical system [82, 116, 117]. A high level overview of the staged sequential procedure for systematic quantitative design of the exoskeleton is shown in Figure 58.

![Figure 58. Staged sequential procedure for knee exoskeleton design](image)

![Figure 59. Schematic of knee exoskeleton](image)
The first stage remains determination of DOF and leading direction for the knee, the exoskeleton and the coupled system. In our study, 6 degrees of knee motion and loading are considered for exoskeleton design. As shown in Figure 59, exoskeleton consists of actuation- and fixation- subassemblies for sequential analysis in our study. Hence we need to apply the methods shown in [104, 105] separately to determine two possible coordinate solutions emerge from application of this methodology are A) nL₁ = 4 (and nR₁ = 2) or B) nL₁ = 3 (and nR₁ = 3) where nL₁ is DOF of fixation subassembly and nR₁ is DOF of actuation subassembly.

Thus, from Jarrasse et al, the optimal DOF of the actuation subassembly could either be (nR₁ = 2) or (nR₁ = 3). Choosing nR₁ = 2 has the advantage of minimalism in terms of actuation requirements. Benefits include reducing number of actuators (and hence weight) and reduced coordination requirements between the multiple actuators but potential drawbacks could increase individual actuation force requirements.

Instead, we prefer to choose the case nR₁ = 3 where the actuation subassembly composes of 3 joints (combination of prismatic/revolute). Within this case, a number of configurations for the knee exoskeleton become possible. Each joint could represent 6 possible motions (Rx, Ry, Rz, Px,
P_y or P_z). Hence 216 (6×6×6) configurations become possible by cascading 3 joints in series. However, not every configuration is feasible (or desirable) and hence we go through a process of elimination to downselect the number of preferred configurations. For example, the space between upright and shank is limited so it is hard to find hardware satisfying both conditions (low-profile and reasonable force generation). This allows us to remove prismatic motion in x-axis (P_x) from active consideration.

Similarly the major rotational knee joint motion is R_x so any motion in other direction causes undesirable torque at knee joint and fixation point; R_y and R_z will not be considered in our knee exoskeleton design. Further, any configurations which have repetition of same prismatic joint and symmetric configurations are deleted from candidates, as shown in Figure 60. However, this process described above is still qualitative and hence we now seek to develop and validate a quantitative approach.

5.3 Analysis for Actuation Subassembly

Nullspace:

The actuation subassembly is considered first and use the methodology described in background section to determine the nullspaces of 8 candidate designs (shown in Table 4). The null space of candidates 8 (P_z-P_y-P_z configuration) has four columns and it has only two degree-of-freedom comparing with three degree-of-freedom of other configurations and hence it will be eliminated.

<table>
<thead>
<tr>
<th>Candidate</th>
<th>((J_{FE}^E)^T)</th>
<th>Null spaces</th>
</tr>
</thead>
<tbody>
<tr>
<td>R_x-R_y-R_z</td>
<td>\begin{bmatrix} 0 &amp; -L_2 \cdot C_3 - L_1 \cdot C_23 &amp; L_2 \cdot S_3 + L_1 \cdot S_23 &amp; 1 &amp; 0 &amp; 0 \ 0 &amp; -L_2 \cdot C_3 &amp; L_2 \cdot S_3 &amp; 1 &amp; 0 &amp; 0 \ 0 &amp; 0 &amp; 0 &amp; 1 &amp; 0 &amp; 0 \end{bmatrix}</td>
<td>\begin{bmatrix} 1 &amp; 0 &amp; 0 \ 0 &amp; 0 &amp; 0 \ 0 &amp; 0 &amp; 0 \end{bmatrix}</td>
</tr>
<tr>
<td>R_x-R_y-P_z</td>
<td>\begin{bmatrix} 0 &amp; -(L_1 \cdot C_2 + L_2 + L_3) &amp; L_1 \cdot S_2 &amp; 1 &amp; 0 &amp; 0 \ 0 &amp; -(L_2 + L_3) &amp; 0 &amp; 1 &amp; 0 &amp; 0 \ 0 &amp; 0 &amp; 1 &amp; 0 &amp; 0 &amp; 0 \end{bmatrix}</td>
<td>\begin{bmatrix} 1 &amp; 0 &amp; 0 \ 0 &amp; 0 &amp; 0 \ 0 &amp; 1 &amp; 0 \ 0 &amp; 0 &amp; 1 \end{bmatrix}</td>
</tr>
</tbody>
</table>
Undesirable Fixation Torque:

Torque at the fixation point can cause discomfort to the users and hence we desire that the most distal joint within the serial chain be passively accommodating (i.e., $\tau_3$ in case of a revolute or $F_3$ in case of a prismatic will be set to 0, Figure 60). Torques and forces resulting at the end effector frame for the remaining 7 design using

$$F_t = [(J_{FE}^E)^T]^{-1}\tau$$  \hspace{1cm} (40)$$

This result is now shown in Table 5 and this will also be used to downselect configuration as discussed below.
<table>
<thead>
<tr>
<th>Candidate</th>
<th>Force and torque</th>
</tr>
</thead>
</table>
| 1 \(\text{R}_x\text{R}_y\text{R}_z\) | \[
\begin{bmatrix}
F_y \\
F_z \\
\tau_x \quad \tau_z = 0
\end{bmatrix} = \begin{bmatrix}
\frac{L_2 \cdot \tau_1 \cdot S3 - L_2 \cdot \tau_2 \cdot S3 - L_1 \cdot \tau_2 \cdot S23}{L_1 \cdot L_2 \cdot S2} \\
\frac{(L_2 \cdot \tau_1 \cdot C3 - L_2 \cdot \tau_2 \cdot C3 - L_1 \cdot \tau_2 \cdot C23)}{L_1 \cdot L_2 \cdot S2} \\
0
\end{bmatrix}
\] |
| 2 \(\text{R}_x\text{R}_z\text{P}_z\)   | \[
\begin{bmatrix}
F_y \\
F_z \\
\tau_x \quad \tau_z = 0
\end{bmatrix} = \begin{bmatrix}
\frac{- (\tau_1 - \tau_3)}{L_1 \cdot C2} \\
0 \\
\frac{\tau_2 + (\tau_2 \cdot L_3 - \tau_1 \cdot L_3 - L_2 \cdot \tau_1 + L_2 \cdot \tau_2)}{L_1 \cdot C2}
\end{bmatrix}
\] |
| 3 \(\text{R}_x\text{P}_z\text{R}_y\)   | \[
\begin{bmatrix}
F_y \\
F_z \\
\tau_x \quad \tau_z = 0
\end{bmatrix} = \begin{bmatrix}
\frac{F_2 \cdot S3 - \tau_1 \cdot C3}{L_1 + L_2} \\
\frac{F_2 \cdot C3 + \tau_1 \cdot S3}{L_1 + L_2} \\
0
\end{bmatrix}
\] |
| 4 \(\text{P}_y\text{R}_z\text{R}_y\)   | \[
\begin{bmatrix}
F_y \\
F_z \\
\tau_x \quad \tau_z = 0
\end{bmatrix} = \begin{bmatrix}
\frac{F_1 \cdot S3}{S2} - \frac{\tau_2 \cdot S23}{L_2 \cdot S2} \\
-\tau_2 \cdot C23 + F_1 \cdot L_2 \cdot C3 \\
\frac{L_2 \cdot S2}{L_2 \cdot S2}
\end{bmatrix}
\] |
| 5 \(\text{R}_x\text{P}_z\text{P}_y\)   | \[
\begin{bmatrix}
F_y \\
F_z \\
\tau_x \quad \tau_z = 0
\end{bmatrix} = \begin{bmatrix}
0 \\
F_2 \\
\tau_1 - F_2 \cdot L_3
\end{bmatrix}
\] |
| 6 \(\text{P}_y\text{R}_x\text{P}_y\)   | \[
\begin{bmatrix}
F_y \\
F_z \\
\tau_x \quad \tau_z = 0
\end{bmatrix} = \begin{bmatrix}
0 \\
-\frac{F_1}{S2} \\
\tau_2 + \frac{F_2 \cdot L_3}{S2}
\end{bmatrix}
\] |
| 7 \(\text{P}_x\text{P}_y\text{R}_y\)   | \[
\begin{bmatrix}
F_y \\
F_z \\
\tau_x \quad \tau_z = 0
\end{bmatrix} = \begin{bmatrix}
\frac{F_2 \cdot S3 - F_1 \cdot C3}{F_2 \cdot C3 + F_1 \cdot S3} \\
0 \\
0
\end{bmatrix}
\] |

Table 5. Force and torque at end effector of actuation subassembly
(Bold indicates acceptable Configuration)

As shown in Table 5, the configuration which has revolute joint at the end does not produce torque at the end-effector.
Singularity:

Similarly singularity of the linkage are undesirable- and can be used a criterion to eliminate certain configuration. The columns of Jacobian(J) can be interpreted as the vector-fields spanning the distribution of feasible twists. The rank of Jacobian of R-R-R, R-P-R, P-R-R and P-P-R configurations are three and the determinant of the Jacobian of four remaining configurations are in Table 6.

<table>
<thead>
<tr>
<th>Candidate</th>
<th>Determinant of Jacobian</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 (\text{R}_x\cdot\text{R}_x\cdot\text{R}_x)</td>
<td>(\text{Det}(J_{FE}^{\text{R}}) = L_1 \cdot L_2 \cdot \sin\theta_2)</td>
</tr>
<tr>
<td>3 (\text{R}_x\cdot\text{P}_x\cdot\text{R}_x)</td>
<td>(\text{Det}(J_{FE}^{\text{R}}) = -L_1 - L_2)</td>
</tr>
<tr>
<td>4 (\text{P}_y\cdot\text{R}_x\cdot\text{R}_x)</td>
<td>(\text{Det}(J_{FE}^{\text{R}}) = L_2 \cdot \sin\theta_2)</td>
</tr>
<tr>
<td>7 (\text{P}_x\cdot\text{P}_y\cdot\text{R}_x)</td>
<td>(\text{Det}(J_{FE}^{\text{R}}) = -1)</td>
</tr>
</tbody>
</table>

Table 6. Determinant of Jacobian (Bold indicates acceptable Configuration)

The singular configuration of both \(\text{R}_x\cdot\text{R}_x\cdot\text{R}_x\) and \(\text{P}_y\cdot\text{R}_x\cdot\text{R}_x\) occurs when \(\theta_2 = 0^\circ, 180^\circ, 360^\circ \ldots (= n\pi, n \in \mathbb{Z})\). Hence, arbitrary motion at the end-effector cannot be passively accommodated at these configurations.

Symmetry:

Symmetry is a highly desirable characteristic for knee brace design – since knee brace loads the knee and vice-versa. Comparing \(\text{P}_z\cdot\text{P}_y\cdot\text{R}_x\) and \(\text{R}_x\cdot\text{P}_y\cdot\text{P}_z\) linkage, \(\text{R}_x\cdot\text{P}_y\cdot\text{P}_z\) cannot provide enough degree of freedom when knee joint back drives knee brace (which causes failure of brace structure or damage to knee joint). Hence in our remaining candidates, we use this argument to help further eliminate non-symmetric configurations.
As the last step in this process, we now combine multiple criteria – the number of degree-of-freedom, undesirable torque at the end-effector (fixation point), singularity and symmetry – together to downselect configurations. The optimal configuration that satisfies all condition is $R_x-P_z-R_x$ linkage which can apply force to $z/y$-axis and allows rotational freedom at the end-effector by setting last revolute joint as frictionless free passive. Figure 61 pictorially depicts the sequential decision process for actuation subassembly and Figure 62 is the configuration of optimal actuation part.

Figure 61. Decision process for optimal design of knee brace active subassembly
5.4 Analysis for Fixation Subassembly

To avoid an occurrence of hyperstaticity and/or uncontrolled interaction force/torque, knee exoskeleton has to be 6 degree–of–freedom mechanism so fixation subassembly should have three degrees-of-freedom. As in the actuation subassembly case, there are multiple candidates for the configuration of fixation subassembly of the 216 combinations of configurations.

By excluding undesirable directions, we can decrease number of candidates as shown in Figure 63. Prismatic joints directed to y and z axis make force generated by actuation subassembly invalid and revolute joint in the direction of x axis is repetition of previous joint, R₃. Thus the feasible joint types and directions for fixation subassembly are Pₓ, Rᵧ and Rᵦ as shown in Figure 64.

Successive 2(3) DOF revolute joints can be realized as universal (or socket) joints respectively which offers a compact and lightweight deployment options. In knee brace design, the space for fixation subassembly is very limited and weight of subassembly is one of crucial
design factor and hence $P_x$-$R_y$-$R_z$ ($P_x$-$R_z$-$R_y$) and $R_y$-$R_z$-$P_x$ ($R_z$-$R_y$-$P_x$) configuration are preferred in fixation subassembly.

![Diagram](image1)

**Figure 63. Undesirable configurations of prismatic and revolute joints for fixation subassembly**

![Diagram](image2)

**Figure 64. Feasible combination of joint configuration**

In our study, we choose $P_x$-$R_y$-$R_z$ ($P_x$-$R_z$-$R_y$) configuration for the fixation subassembly because compact socket joint is not usually commercially available and its workspace is quite limited comparing with universal joint.

### 5.5 Analysis for Complete Knee Brace

Figure 65 shows the final knee brace design featuring the $R_x$-$P_z$-$R_x$ actuation subassembly and the $P_x$-$R_y$-$R_z$ fixation subassembly. As a quick final check, we perform the twist- and
wrench-analysis for the entire knee brace and compute the Torque and force at the end-effector \( \{E_2\} \)

\[
\begin{bmatrix}
F_x \\
F_y \\
F_z \\
\tau_x \\
\tau_y \\
\tau_z
\end{bmatrix}
=\begin{bmatrix}
f_{x1} \cdot \tau_1 + f_{x2} \cdot F_2 + f_{x3} \cdot \tau_3 + f_{x4} \cdot F_4 \\
f_{y1} \cdot \tau_1 + f_{y2} \cdot F_2 + f_{y3} \cdot \tau_3 + f_{y4} \cdot F_4 \\
f_{z1} \cdot \tau_1 + f_{z2} \cdot F_2 + f_{z3} \cdot \tau_3 + f_{z4} \cdot F_4 \\
\tau_3 \cdot C6 + \tau_5 \cdot C5 \cdot S6 - \tau_6 \cdot C6 \cdot S5 \\
-\tau_3 \cdot S6 + \tau_5 \cdot C5 \cdot C6 + \tau_6 \cdot S5 \cdot S6 \\
\end{bmatrix}

\[
\frac{C5}{\tau_6}
\]

By assuming that the first two joints of actuation subassembly are active and the rest of them are frictionless passive joints, the torques and forces at fixation point, \( \{E_2\} \) are now:

\[
\begin{bmatrix}
F_x \\
F_y \\
F_z \\
\tau_x \\
\tau_y \\
\tau_z
\end{bmatrix}_{\tau_3, F_4, \tau_5, \tau_6=0}
=\begin{bmatrix}
f_{x1} \cdot \tau_1 + f_{x2} \cdot F_2 \\
f_{y1} \cdot \tau_1 + f_{y2} \cdot F_2 \\
f_{z1} \cdot \tau_1 + f_{z2} \cdot F_2 \\
0 \\
0 \\
0
\end{bmatrix}

Thus this verifies that the final configuration, \( R_x-P_x-R_x-P_y-R_z \) is able to apply forces perpendicular to the axis of limb without creating undesirable moment at fixation point.

Figure 65. Configuration of (a) actuation (Sagittal plan) and (b) overall Subassembly
5.6 Physical Verification for Knee Exoskeleton

Figure 65 shows schematics of final configuration of knee exoskeleton which is $R_x$-$P_y$-$R_x$-$P_y$-$R_y$-$R_z$. The first revolute and second prismatic joints are active and the rest of them are free passive joints.

**3D Printed Scaled Active Knee Brace:**

To verify our final knee exoskeleton design, 3D printed scaled active knee brace was developed. Prismatic and revolute joint are actuated by servo motors, the force and torque at fixation point are measured by 6-DOF force transducer (ATI Nano17) and a universal joints substitutes for two successive revolute joints as shown in Figure 66.

![Figure 66. Scaled Active Knee Brace Prototype](image)

The main purpose of scaled knee brace test is to measure the level of undesired force and torques which provides discomfort to users. Revolute ($R_1$) and prismatic ($P_2$) joints are activated...
from zero to full power level and load at fixation point, $\{E_2\}$ is measured by 6-DOF force transducer with respect to various configurations of knee.

![Figure 67. Force and Torque at Fixation Point ($\{E_2\}$) when (a) revolute joint ($R_1$) (b) prismatic joint ($P_2$) is activated separately](image)

The final design configuration is evaluated by 3D printed scaled knee exoskeleton test. As shown in Figure 67, undesired force ($F_x$) and torque ($\tau_x, \tau_y, \tau_z$) are recorded which mainly caused by friction/tolerance and moment arm between location of sensor and center of joint. The maximum torque measured at fixation point is 0.015Nm which negligibly smaller than forces (Max. 6N) as depicted by black lines (torques) in Figure 67.

We examine the use of screw-theoretic analysis tools to provide a systematic and quantitative framework for design, formulation and evaluation of knee exoskeleton. Specifically we emphasize on automatic and systematic design process which can be easily extended to upper limb or whole body exoskeleton. In ideal condition, with light weight and frictionless parts, this method allows us to design knee exoskeleton to be selectively isolated. It applies force perpendicular or parallel to axis of limb and does not transmit undesirable torque to the user.
6  Smart Brace Design - Compliant Mechanism Design

6.1 Rigid and Flexible Exoskeleton

![Image](a)

![Image](b)

Figure 68. (a) Rigid Type Exoskeleton, HAL from Cyberdyne [99] and (b) Cable-Driven Arm Exoskeleton (CAREX) [118]

Our broad research goal is to pursue the design of exoskeletons/braces that use passive and semi-active accommodation and redirection of the force to achieve their function. The standard manipulator design paradigm assumes a rigid link manipulator with all mobility restricted to the joints as seen in the prototypical HAL exoskeleton in Figure 68-(a). The links of manipulator can usually be stiffened by adding structural mass or by improving the material properties but this will increase the total mass of the system which is not generally acceptable to exoskeleton design [119]. Cable-driven exoskeletons allows us to move actuators to the base from joint which can reduce the mass and inertia of moving bodies as seen in the CAREX design [118, 120]. The main limitations of cable-driven system, however, arise from the tension-only capability of cables and need for using n+1 cables to realize control of an n-DOF manipulator.

Recent designs have introduced flexibility at either the link or directly at the joint. In particular, there has been a movement in the mechanical design community to further a paradigm
of compliant mechanisms [121]. Compliant mechanisms are a class of multibody articulated systems that achieve transfer of power (motion and force) through elastic deformation [122]. Such elastic deformation could be localized at the joints or distributed throughout the links. Many examples of compliant mechanisms have been developed, using a rigid-body-replacement-synthesis process. However, the use of compliant mechanisms to serve as flexural elements for mesoscale systems including exoskeletons/braces have not been widely explored (with notable exceptions such as [122-127]) due to the significant load requirements.

6.2 Design Overview

![Diagram showing design objective]

Figure 69. Design Objective

We examine the development of a hybrid lightweight compliant/rigid mechanism, with a passive/semi-active means to adjust stiffness and customize the load while bracing the user’s knee. The Parallel Coupled Compliant plate (PCCP) mechanism and Pennate Elastic Band (PEB) spring system introduced in this dissertation, provides both flexibility and extreme stiffness to user with respect to posture/angle of knee joint, without mediation of active devices and even
active sensors. In addition to passive mode of operation of the PCCP/PEB system, semi-active type design is exploited to allow for changes of preload of PEB spring to provide force/torque customization capability.

6.3 Parallel Coupled Compliant Plate (PCCP) Mechanism

The Parallel Coupled Compliant Plate (PCCP) represents a hybrid mechanism which combines advantages of rigid and flexible link/joint for knee exoskeleton.

![Prototype of Parallel Coupled Compliant Plate (PCCP) Mechanism](figure70)

Figure 70. Prototype of Parallel Coupled Compliant Plate (PCCP) Mechanism, video is available from Youtube (www.youtube.com/watch?v=wPuYBphedz8)

As shown in Figure 70, PCCP mechanism consists of base/sliding plate, sliding pins/slots and fixed pins. Two parallel compliant plates are fixed with pins and all pins are permanently fixed to lower plate. The motion of sliding pins is constrained and limited by sliding slots. Fixed pins located at the center of plate make sectional deformation contour symmetric. The shape of sectional contour can be modified by changing the location of fixed pins. The shape, length, the location of fixed pins and pattern of sliding slot can be customized for desired sectional contour of PCCP.
As shown in Figure 71, when PCCP is bending, sliding pins are moving freely in the inside of sliding slot as shown in Figure 71-(a) and motion of pin is guided by the shape of sliding slots.

![Figure 71. (a) Beam bending mode and (b) Fixed-guided bending mode](image)

In this mode, the deformation of PCCP is similar to the bending of cantilever beam (beam bending mode). The ideal sectional contour of plate is considered as a shape of arc because all sliding pins are deployed evenly and keep the normal distance between base and sliding plate constant. Fixed-guided bending mode which increases the stiffness of PCCP drastically, starts when the motion of sliding pin is limited by sliding slot as shown in Figure 71 – (b).

### 6.4 Compliant Mechanism Theory

Conventional analysis method for compliant mechanism is particularly useful when a design has been chosen and the geometry, material properties and load conditions are available. The pseudo-rigid-body model concept is used to model the deflection of flexible mechanism using rigid-body components that have equivalent force-deflection characteristics. The motion is modeled by rigid links connected with revolute joint. Springs are added to the joint to accurately predict the force-deflection relationships of the compliant mechanism. Thus, the pseudo-rigid-
body model provides a simple method to analyze systems, and predict the deflection path and force-deflection relationships. However, carefully determining the pivot location and effective nonlinear spring constant is critical to its successful deployment.

Consider a flexible cantilever beam with constant cross section and linear material properties as shown in Figure 72. A flexible cantilever beam with a force at the free end, the free end follows a nearly circular path and this idea is used for approximation of the beam’s deflection path.

![Figure 72. Cantilever beam with force at the free end and its pseudo-rigid-body model](image)

Figure 72-(b) shows a pseudo-rigid-body model of a large deflection beam which is modeled by two rigid links joined at a pivot along the beam. A torsional spring at the pivot represents the beam’s resistance to deflection. The optimal value of the characteristic radius factor, $\gamma$ is decided by optimization process as following.

Maximizes the pseudo-rigid-body angle, $\Theta$

$$\Theta = \arctan \frac{b}{a - l(1 - \gamma)}$$  \hspace{1cm} (41)

Subject to
\[ g(\theta) = \frac{error}{\delta_e} \leq \left[ \frac{error}{\delta_e} \right]_{\text{max}} \] (42)

Where \( \frac{error}{\delta_e} \) is relative deflection error and \( \delta_e = \sqrt{(1-a)^2 + b^2} \)

Then optimal value of the characteristic radius factor is

\[
\gamma = \begin{cases} 
0.841655 - 0.0067807n + 0.000438n^2 & (0.5 < n < 10.0) \\
0.852144 - 0.0182867n & (-1.8316 < n < 0.5) \\
0.912364 + 0.0145928n & (-5 < n < -1.8316) 
\end{cases}
\] (43)

The resistance of beam is modeled by a torsional spring, \( K_\theta \), called stiffness coefficient. Combined with geometric and material properties, the stiffness coefficient is used to determine the value of the spring constant for a particular beam’s pseudo-rigid-body model as following.

The stiffness coefficient, \( K_\theta \), can be approximated as

\[
K_\theta = \begin{cases} 
3.024112 + 0.121290n + 0.003169n^2 & (-5 < n < -2.5) \\
1.967647 - 2.616021n - 3.738166n^2 - 2.649437n^3 - 0.891906n^4 & (-2.5 < n < -1) \\
-0.113063n^4 & (-1 < n < 10) \\
2.654855 - 0.509896 \times 10^{-1}n + 0.126749 \times 10^{-1}n^2 - 0.142039 \times 10^{-2}n^3 + 0.584525 \times 10^{-4}n^4 & (-1 < n < 10) 
\end{cases}
\] (44)

then torsional spring constant, \( K \) is [121],

\[ K = \gamma K_\theta \frac{E l}{l} \] (45)

### 6.5 Pseudo Rigid Body Model

PCCP mechanism introduced in Figure 70 has various design parameters to be decided including position/number of pins, patterns of the sliding slots, material properties and dimension of plate which can be decided by loading and desired kinematic configuration of the mechanism.
Table 7. Schematic and Pseudo-rigid-body model for (a) beam bending mode and (b) fixed-guided bending mode

PCCP has two distinct modes called beam bending mode and fixed-guided bending mode which can be decided by the location of sliding pin in sliding slot as depicted in Table 7. *Beam bending mode* is similar with a cantilever beam loading at the free end and can be accurately modeled by two rigid links that are joined at a pivot. A torsional spring at the pivot represents the beam’s resistance to deflection. *Fixed-guided bending mode* is a loading condition that one end of the beam is fixed while the other is guided which can be modeled by three links connected with two revolute joint and loaded by torsional springs.

### 6.6 Modes of PCCP Mechanism

#### Beam Bending Mode

As discussed in previous section, PCCP shows two distinctive modes - *beam bending and fixed-guided bending mode.*
Figure 73. Beam bending mode and fixed-guided bending mode of PCCP

At the excessive extension/flexion knee posture/angle, PCCP is at fixed-guided bending mode and protects knee joint by changing the stiffness of structure into extreme level. On the other hand, PCCP can provide predesigned assistive load to knee joint for better mobility of users at normal range of knee angle as shown in Figure 73.

Figure 74. Bending Model of PCCP Mechanism for Knee Exoskeleton

Symmetric configuration of PCCP and evenly distributed pins - spanning two flexible plates, constrain the shape of structure and sustain sectional contour of PCCP as a shape of arc.
Thus we assume the ideal sectional contour of flexible plate as an arc whose center and radius are changing with respect to the deformation/bending of the mechanism. The deflection of base plate is calculated by geometry as shown in Figure 74.

\[ x_1 = R_1 \sin \alpha_1, \quad y_1 = R_1 - R_1 \cos \alpha_1, \quad (R_1 = L_0/2\alpha_1) \]  \hspace{1cm} (46)

And the deflection of sliding plate \((x_2, y_2)\) is

\[ x_2 = R_2 \sin \alpha_2, \quad y_2 = R_2 - R_2 \cos \alpha_2 \]  \hspace{1cm} (47)

Where \(R_2 = R_1 + g\) and \(\alpha_2 = L_0/2R_2\)

Figure 75 shows a pseudo-rigid-body model (one link mechanism) of a beam bending mode of PCCP. A torsional spring at the pivot represents the linkage’s resistance to deflection.

![Figure 75](image)

Figure 75. (a) Pseudo-rigid-body Model of PCCP (beam bending mode) and (b) Deflection of PCCP - Ideal bending model vs. pseudo-rigid-body model

The location of characteristic pivot is defined by \(L_0(1 - \gamma)\) and characteristic radius is the radius of the circular deflection path which is \(\gamma L_0\) where \(\gamma\) is characteristic radius factor [121]. The position of end-effector is simply:
\[
\begin{bmatrix}
\dot{x} \\
\dot{y}
\end{bmatrix} = \begin{bmatrix}
L_0(1 - \gamma) \\
0
\end{bmatrix} + \begin{bmatrix}
\gamma L_0 \cos \theta \\
-\gamma L_0 \sin \theta
\end{bmatrix}
\] (48)

And the reliability of pseudo-rigid-body model can be proven by plotting the deflection of ideal bending model and pseudo-rigid-body model (Figure 75). An optimal value for the characteristic radius factor, \(\gamma\), is found by optimization search to minimize the error between trajectory of ideal bending and pseudo-rigid-body model. PCCP (at beam bending mode) can be approximately modeled by cantilever beam when the motion of slider pin is not limited by guiding slot. The deflection of cantilever beam with a force at the free end is calculated approximately as following. Consider a flexible stainless beam (\(E = 200\) GPa, STS304) that is 310mm long and has rectangular cross section – a width of 25mm and a thickness 0.5 mm (dimension for a plate of PCCP prototype). The torsional spring constant for pseudo-rigid-body model is:

\[
K_{Bending} = \gamma K_\Theta \frac{EI}{I}
\] (49)

Where \(\gamma\) is characteristic radius factor, \(l\) is length of beam, \(K_\Theta\) is stiffness coefficient, \(E\) is young’s modulus and \(I\) is the moment of inertia [121].

Figure 76. Force at the free end and torque at characteristic pivot of pseudo-rigid-body model (Single cantilever beam, \(L=310\text{mm}, W=25\text{mm}, t=0.5\text{mm}, E=200\text{ GPa}\))
As shown in Figure 76, maximum torque at characteristic pivot is 0.3 Nm and maximum normal force at free end is 2.25N (for 45 degree bending) which is not enough for knee exoskeleton application considering the magnitude of loading at knee joint as discussed in introduction section. A unique characteristic – arc length difference between upper and lower plate during deformation – allows PCCP exoskeleton to provide increased and customized force profile to the user by combining with Pennate Elastic Band (PEB) spring.

*Fixed-guided Bending Mode*

The stiffness of PCCP increases discontinuously and drastically when the motion of sliding pins are limited by sliding slots and hard to move further as shown in Figure 77. This fixed-guided mode can be modeled by a pseudo-rigid-body model which one end of beam is fixed while the other is guided with fixed angle. To maintain the constant angle, a resultant moment has to be present at the end.

![Figure 77. (a) Fixed-guided bending of PCCP and (b) Pseudo-rigid-body model of PCCP](image)

The length of unit segment of PCCP prototype is 33mm and it consists of 8 unit segments. Those are distributed symmetrically from the center of PCCP as shown in Figure 77. For each of
two springs, the torsional spring constant, $K$ is twice as stiff as for the case of cantilever beam and there are two springs for the pseudo-rigid-body model. Thus the unit segment is four times as stiff as a cantilever beam with same length [121]. For our PCCP prototype, it can be considered as rigidly connected 8 fixed-guided segments. The torsional spring constant for pseudo-rigid-body model is

$$K_{Fixed} = \frac{n}{2} \times 4 \times \gamma K_\theta \frac{EI}{I_{unit}} \quad (50)$$

As shown in Figure 78, at fixed-guided bending mode, PCCP can support much higher load than the case at beam bending mode. It can support 22N (normal) at 5 degree of deflection. This unique property of PCCP can provide exclusive opportunity in knee exoskeleton design which is absent in conventional mechanisms. The stiffness of PCCP knee exoskeleton can be changed drastically when user’s knee joint angle or posture reaches dangerous region which causes serious injury to knee joint.

Figure 78. Resultant force, $P$ of PCCP (Eight unit segments, Fixed-guided bending mode, Unit length – 33mm, Total length – 264mm)
6.7 Pennate Elastic Band (PEB) Spring

To increase and control the stiffness of PCCP at beam bending mode, we developed Pennate Elastic Band (PEB) spring. Bio-inspired (pennate muscle, Figure 79-a) PEB spring consists of base and slider part which are connected to the compliant plates of PCCP separately. The base and slider part are connected by a prismatic joint. The arc length difference of base/sliding plates of PCCP caused by bending generates the relative motion between base and slider part of PEB spring which stretches pennate elastic bands as shown in Figure 79-b. Obliquely attached multiple elastic bands structure allows higher force production with smaller range of motion and size.

The force profile of PEB spring can be customized by number, pennation angle and superposed geometric pattern of elastic band. In this paper we only use simple straight pattern and the detail of PEB spring design will be discussed in consecutive paper. Figure 80 – (a) shows

Figure 79. (a) Analogy between Pennate muscle (Courtesy of McGraw-Hill) and PEB spring and (b) Prototype of Pennate Elastic Band (PEB) Spring with PCCP

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schematic of basic pattern of PEB spring and force profile with respect to displacement of sliding part (d) in Figure 80–(b). The resultant force \( F_n \) of PEB spring can be defined by:

\[
F_n = 2nk \left[ 1 - \frac{p}{\sqrt{d^2 + p^2}} \right]
\]  

(51)

where \( n \) is number of elastic bands, \( \beta \) is pennation angle, \( F_p \) is tension- and \( k \) is spring constant- of unit elastic band.

![Schematic of PEB spring and test setup](image1)

![Resultant force of PEB with respect to joint angle of pseudo-rigid-body model](image2)

Figure 80. (a) Schematic of PEB spring and test setup and (b) Resultant force of PEB with respect to joint angle of pseudo-rigid-body model (\( n=10 \), \( p=14 \text{mm} \), \( k = 0.3 \text{N/mm} \), dimension from PEB spring prototype)

PCCP combined with two PEB spring can be modeled as a single link mechanism (with a torsional spring located at characteristic pivot) and a nonlinear spring fixed to the ground as depicted in Figure 81.

The linear elongation of nonlinear spring, \( k_{nl} \) is

\[
dl_{nl} = \sqrt{\left( (L_0 + l_0) - x \right)^2 + y^2} - l_0 = \left( \sqrt{2(l_0^2 y^2 + l_0 L_0 y)(1 - \cos \theta)} + l_0^2 \right) - l_0
\]  

(52)
Figure 81. (a) Pseudo-rigid-body model of PCCP/PEB system and (b) Resultant force of PEB spring and nonlinear spring model with respect to joint angle of pseudo-rigid-body model (n=10, p=21mm, k = 0.3N/mm, g=40mm)

Arc length difference between base plate and slider plate projected to base plate is

$$\Delta Arc = Arc_{base} - Arc_{slide} = R_1(\alpha_1 - \alpha_2) = R_1 \left( \alpha_1 - \frac{L_0}{2(R_1 + g)} \right)$$ \hspace{1cm} (53)

Optimal length of nonlinear spring (approximation model for PEB spring connected to PCCP) can be decided by finding $l_0$ which satisfies the relationship as following

$$R_1 \left( \alpha_1 - \frac{L_0}{2(R_1 + g)} \right) = \sqrt{2(L_0^2 \gamma^2 + l_0 l_0 \gamma)(1 - \cos\theta) + l_0^2} - l_0$$ \hspace{1cm} (54)

Figure 81-b shows spring force with respect to the joint angle ($\theta$) of PEB spring and the nonlinear spring of pseudo-rigid-body model in Figure 81- (a).

6.8 Semi-active Design for Customized Load Application
Figure 82. Prototype of semi-active PCCP/PEB system (a) CAD model and (b) 3-D Printed Prototype

Figure 82 shows CAD model and prototype of semi-active PCCP/PEB system. The preload adjustor located between base and sliding part adjusts the preload of PEB spring which allow us to modify force/torque profile generated by PCCP/PEB system in detail for better customization.

Figure 83. (a) Pseudo-rigid-body model of Semi-active PCCP/PEB system and (b) Spring force of semi-active PEB spring

The linear elongation of nonlinear spring with preload adjustor, $d_{nlp}$ is

$$dl_{nlp} = dl_{nl} - dl_t = \left( \sqrt{2(L_0^2 \gamma^2 + L_0 \gamma (1 - \cos \theta)) + l_0^2} \right) - l_0 - dl_t$$  \hspace{1cm} (55)
The displacement of preload adjustor, $dl_{\text{nlp}}$, can be any continuous functions which is able to be generated by a linear actuator. As noted previously, the preload adjustor can be controlled by a linear-actuator along various profiles. Figure 83 depicts (a) the pseudo-rigid-body model and (b) theoretical spring forces of semi-active PEB spring for various preload-adjuster/linear-actuator input-profiles. The three Semi-Active Mode curves correspond to the force generated by the PEB in the semi-active mode when the preload adjuster (linear actuator) is moved as (i) cosine (dotted), (ii) sine (dashed) or (iii) linear proportional (dash-dot) function of the joint angle. (iv) The solid line denotes the force generated by the PEB in the passive mode (when the linear actuator/preload adjuster is deactivated).

6.9 Design Verification with Physical Prototype

![Prototype Image]

**Figure 84. Full scaled PCCP/PEB prototype test with saw-bone knee model**

The performance of full scaled PCCP/PEB system is tested by motion capture camera and 6-DOF force transducer as shown in Figure 84. The spatial motion of saw-bone knee model is
measured by motion capture camera (OptiTrack V100:R2 [128]) which is tracking reflective markers attached on anatomical landmarks of femur and tibia bone in real time (100Hz). The force and moment during flexion/extension motion are measured by 6-DOF force transducer (ATI Delta [92]) which is rigidly connected to tibia bone of knee model.

First, we measured the load at knee joint without PCCP/PEB system and repeated same process (flexion and extension motion) with knee exoskeleton as shown in Figure 85.

![Figure 85. (a) Test with knee model only and (b) with PCCP/PEB exoskeleton](image)

The contribution of PCCP/PEB exoskeleton (assistive force and torque) is distinctively shown in Figure 86 (b). Dotted red lines are load at knee joint without exoskeleton and blue solid lines are load with exoskeleton.

The force-deflection curve is $C^0$ continuous but a change of slope occurs as the mode switches from beam-bending mode (Mode A) to fixed-guided bending mode (Mode B). Figure 86 - (a) reflects this significant stiffness-transition that can protect knee against hyperextension.
Figure 86. Torque at knee joint (a) at beam bending and fixed-guided bending mode, (b) with/without knee brace

Figure 86 - (b) shows the positive assistive-torque with increasing joint-angle that aids the flexion motion of saw-bones model operated with/without the knee-brace (now solely operating in beam-bending Mode A). We have attempted to carefully eliminate sources of stick-slip friction in the capture of the torque-assist/deflection data.

Figure 87. (a) Stiffness test in fixed-guided bending mode for flexion (right) and extension (left), (b) force-x at knee rotation center for extension motion

The stiffness at fixed-guided bending mode is also tested as shown in Figure 87. Red arrow (Figure 87 - (a)) depicts the direction of resistive stiffness and femur part is pushed to the direction of hyperextension motion, which can cause serious knee injury. Figure 87 - (b) shows
estimated and measured moment at knee joint. The difference between estimated and measured value is mainly caused by accumulated tolerance of prototype, resolution of optical tracking system, friction and deformation of 3D printed parts.

At beam bending mode, PCCP/PEB system provides assistive forces and torques at knee joint contributing to protect knee from injury and increase mobility of user. The stiffness of PCCP/PEB system at beam bending mode can be modified by number/preload of elastic bands but we limited the maximum force from PEB spring to protect test bed especially, force transducer.

An early prototype of a wearable PCCP/PEB Smart Knee Brace – weighing 500g for passive and 1kg for semi-active prototype – is shown in Figure 88. In addition to lightweight advantage, PCCP/PEB system-we developed, offers new opportunity to the design of knee exoskeleton which satisfy the needs of flexibility in safe- and extreme stiffness in excessive- range of knee motion.

![Figure 88. Wearable prototype of (a) passive and (b) semi-active PCCP/PEB knee brace](image)
7 Discussion

Home-based rehabilitation prototype integrated a measuring system, smart knee brace and host system. The measuring system comprised both the Kinect and Wii Balance Board to record both kinematic and static data of participants. The Smart knee brace consisted of compliant mechanism, servo motor, IMU sensor and microcontroller which communicate with the host system. The host system implemented by Matlab GUI, records subject’s motion data transmitted from measuring system and analyzed it to modify parameters of smart knee brace. By changing parameters, smart knee brace provided a customized force/torque profile to the users which are based on current measurement and prescription by therapist, as shown in Figure 89.

![Figure 89. Home-based Rehabilitation System Scenario](image-url)
During the calibration process either in laboratory or clinic, all kinematic information (including marker and sensor position) is collected and saved into a calibration file and later transferred to the host PC in the patient’s house, as shown in Figure 89.

![Calibration setup and home measurement setup for motion capture](image)

**Figure 90.** (a) Calibration setup and (b) home measurement setup for motion capture consists of optical tracking system, Kinect, Wii Balance Board

After calibration (in laboratory and (or) clinic), patients recorded their motion using the low-cost sensor (Kinect) and pc-based application as shown in Figure 91 – (b). Kinematic and quasi-static data captured by Kinect sensor and Wii balance board were transmitted to the host system, which then calculated knee load profiles during quasi-static/dynamic trails, then desired force/torque profile based on pre-programed prescription or by therapist can be accessed remotely.
Figure 91. GUI of host system for home measurement

The semi-active smart knee brace that was introduced in section 4.5 has the capability of customizing force/torque profiles in the similar level of fully-actuated exoskeletons with less power consumption, simpler and lighter configuration. The host system that transfers the desired loading profile to the smart knee braces, minimizes the error between the actual and desired knee force/torque.

Figure 92. (a) CAD model and (b) prototype of semi-active PCCP/PEB smart knee brace
To date, we have successfully prototyped a home-based rehabilitation system for persons with orthopedic dysfunction, which consists of hardware, software and protocols. Two motion capture systems – a low-resolution and low-cost Kinect system and the more-expensive, higher fidelity Vicon motion capture system – were examined to aid quantitative lower limb motion estimation in a clinically-focused squatting study. As expected, the Vicon system with AMS/Visual-3D post-processing yielded outstanding results (with huge workspace, high resolution and sampling rate). Direct application of the Kinect system to a clinical or research setting (without post-processing of raw data) however is limited. However, we noted that with suitable post processing offers potential for applicability for clinically relevant use.

On human identification with Kinect sensor, our operant hypothesis was that the geometric consistency of human musculo-skeletal system and statistical averaging would allow exploitation of the Kinect data to realize considerable accuracy and consistent performance. This is examined in detail in our work to determine subject-geometric-parameters and reconfirmed by the subject classification rate. Classifications rates approached 95.6% on average for data recorded on four separate days and different time to maximize instability of subject physical condition.

Second, kinetostatic and screw-theoretic analysis tools were used to provide a systematic and quantitative framework for design, formulation and evaluation of knee exoskeleton. Specifically emphasis focused on the automatic and systematic design process which then can be easily extended to upper limb or whole body exoskeleton. Brace design process today mostly depends on expert’s experience, knowledge and intuition that are not easily quantifiable. The quantitative knee-bracing design, espoused in this work, seeks to address this limitation by customizing parameters of the knee brace subject to design objectives.
The PCCP and PEB spring design was tested by full-scaled physical prototype. The system was modeled by a pseudo-rigid-body framework, traditionally used for modeling compliant mechanisms. The developed model can predict performance of the PCCP/PEB. In addition, a prototype system was developed and performance of a saw bone model (with/without PCCP/PEB exoskeleton) was measured using an optical tracking system and 6-DOF force transducer.

At beam bending mode, PCCP/PEB system provides assistive forces and torques at knee joint contributing to protect knee from injury and increase mobility of user. The stiffness of PCCP/PEB system at beam bending mode was modifiable by elastic band configuration in PEB spring. The prototype of a wearable PCCP/PEB Smart Knee Brace weighs 500g for passive and 1kg for semi-active prototype. In addition to lightweight advantage, the PCCP/PEB system offers new opportunity to the design of knee exoskeleton which satisfy the needs of flexibility in safe- and extreme stiffness in excessive- range of knee motion.

Our focus in this work was restricted to demonstrating the feasibility of custom-torque assist (along the knee flexion/extension direction). The system is intended to store energy in the compliant-members during knee-flexion phase and release this in the form of the assistive-torque during the transition back to the neutral knee position. This includes: (i) coarse/large magnitude passive-torque-assist using the inherent member-compliance of the PCCP; coupled with (ii) finer/small-magnitude torque-assist customization provided via the passive/semiactive PEB spring system. We limited the maximum force from PEB spring to protect test bed, force transducer and prevent any accident caused by failure of structure. Further, the current analysis does not extend to stress/fatigue testing of the components of this brace – while it is a critical part of any final production design, it is pursuing in our future work. The video of overall
procedure for our Home-based Rehabilitation System for Deficient Knee Patient is available from www.youtube.com/watch?v=WqvO7QMoP3k.
8 References


Appendix

A.1 Motion capture

Marker Set

This marker set is chosen for AMS (Anybody Modeling System) analysis and additional markers will be added for the analysis by Nexus system.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Anybody</th>
<th>Segment</th>
<th>Anybody</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>No.</td>
<td>Name</td>
<td>Upper body</td>
</tr>
<tr>
<td>Lower body</td>
<td>1,2</td>
<td>RTOE/LTOE</td>
<td>23,24 RSHO/LSHO</td>
</tr>
<tr>
<td></td>
<td>3,4</td>
<td>RMT5/LMT5</td>
<td>25,26 RUPA/LUPA</td>
</tr>
<tr>
<td></td>
<td>5,6</td>
<td>RHEE/LHEE</td>
<td>27,28 RELB/LELB</td>
</tr>
<tr>
<td></td>
<td>7,8</td>
<td>RANK/LANK (T1)</td>
<td>29,30 RFRA/LFRA</td>
</tr>
<tr>
<td></td>
<td>9,10</td>
<td>RTIB/LTIB (T2)</td>
<td>31,32 RWRA/LWRA</td>
</tr>
<tr>
<td></td>
<td>11,12</td>
<td>RKNE/LKNE (F1)</td>
<td>33,34 RWRA/LWRA</td>
</tr>
<tr>
<td></td>
<td></td>
<td>13,14 RTHI/LTHI (F2)</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>15,16 RASI/LASI</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>17,18 RPSI/LPSI</td>
<td></td>
</tr>
<tr>
<td>Upper body</td>
<td>19</td>
<td>T10</td>
<td>37,38 RFHD/LFHD</td>
</tr>
<tr>
<td></td>
<td>20</td>
<td>STRN</td>
<td>49,40 RBHD/LBHD</td>
</tr>
<tr>
<td></td>
<td>21</td>
<td>C7</td>
<td>41,42 RTT/LTT (T3)</td>
</tr>
<tr>
<td></td>
<td>22</td>
<td>CLAV</td>
<td>43,44 RGT/LGT (F3)</td>
</tr>
<tr>
<td>Total</td>
<td></td>
<td></td>
<td>44</td>
</tr>
</tbody>
</table>

Markers written in bold are for each three markers of tibia and femur bone. Those markers are used for analysis of relative orientation of femur and tibia.

Example of a subject with attached markers

(Courtesy of CMU Graphics Lab Motion Capture Database)

Comparison Table

(between figure and AMS model)

<table>
<thead>
<tr>
<th>Name in Figure</th>
<th>Name in AMS Model</th>
</tr>
</thead>
<tbody>
<tr>
<td>RFWT</td>
<td>RASI</td>
</tr>
<tr>
<td>LFWT</td>
<td>LASI</td>
</tr>
<tr>
<td>RBWT</td>
<td>RPSI</td>
</tr>
<tr>
<td>LBWT</td>
<td>LPSI</td>
</tr>
<tr>
<td>RLEG</td>
<td>RTIB</td>
</tr>
<tr>
<td>LLEG</td>
<td>LTI</td>
</tr>
<tr>
<td>RARM</td>
<td>RFRA</td>
</tr>
<tr>
<td>LARM</td>
<td>LFRA</td>
</tr>
<tr>
<td>R10 (not use)</td>
<td>N/A</td>
</tr>
</tbody>
</table>
Guide line for placing markers
TT - tibial tuberosity.

C7 - is the most prominent vertebra which can be felt at the base of the neck with the chin resting on the chest. The marker should be placed on this prominence, which is a few centimeters below the hair line.

T10 - is the tenth thoracic vertebra. The precise way to locate this is to ask the performer to lean forward, so that the vertebrae can be felt, and to count 10 vertebrae down from C7.

STRN - is short for Sternum, the breast bone. This marker should be placed on the lower end of the breast bone.

CLAV - should be placed centrally, on the collarbone (or clavicle), just below the throat. This point may tend to be obscured if the head is lowered, but in most movements, it will be clearly visible.

Shoulder - should be placed on top of the shoulder, at a point which remains visible. The bony knob at the end of the collarbone (Acromion) is suitable.

RASI/LASI - The front waist markers should be placed on the bony prominence of the pelvis which are easily felt on either side of the ‘belt buckle’ position.

RPSI/LPSI - The rear waist markers should be positioned relative to the two small dimples found in the ‘small’ of the back. Place the markers on the flat skin just outside the dimples and at the same height.

RKNE/LKNE - This is placed on the outside of the knee at its widest part roughly level with the middle of the kneecap and slightly less than half way between front and back of knee. You can visualize the mechanical flexion of the knee by asking your subject to flex and extend the knee while looking side on.

Pelvis
Foot
Ankle

(Courtesy of CMU Graphics Lab Motion Capture Database)
RANK/LANK - should be close to, or on, the prominent bone of the outside of the ankle.

RTHI/LTHI/RTIB/LTIB - try and have them at slightly different distances from the knee markers. This preserves a sense of anti-symmetry between limbs.

RMT5/LMT5 - the joint of the small toe
RTOE/LTOE - the center of the big toe
RELB/LELB - Outside of the elbow joint, at the widest part of the bone
RFIN/LFIN - on the hand, near the knuckle of the middle finger
RFHD/LFHD - should be placed above the temples
RBHD/LBHD - should be placed diagonally opposite the front head markers

**Kinect System**

Kinect sensor should locate in the center position of the subject (Distance from subject and sensor is about 2670 mm)

![Kinect System Diagram](image)

**Horizontal field of view (0.8m – 4m)**

**Vertical Field of View**

(Courtesy of Microsoft)

**Check points for the position of subjects**

Subject locates in frontal edge of workspace and both feet are in workspace clearly. No subject except main subject should not in workspace it is better to clean every object in workspace.
Quick check list for Kinect sensor

<table>
<thead>
<tr>
<th>Check Items</th>
<th>Check box</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Kinect sensor locates in the center of subject</td>
<td></td>
</tr>
<tr>
<td>2. There is not second subject in the workspace</td>
<td></td>
</tr>
<tr>
<td>3. Subject locates in the frontal edge of work space</td>
<td></td>
</tr>
<tr>
<td>4. Both feet are in the workspace and clearly visible</td>
<td></td>
</tr>
<tr>
<td>5. The base of Kinect sensor is parallel to the ground</td>
<td></td>
</tr>
<tr>
<td>(Edge of workspace is aligned to edge of force plate)</td>
<td></td>
</tr>
<tr>
<td>6. Synchronizer is visible in RGB screen</td>
<td></td>
</tr>
<tr>
<td>7. Kinect sensor faces to the front of subject</td>
<td></td>
</tr>
<tr>
<td>8. No other object in workspace except subject</td>
<td></td>
</tr>
<tr>
<td>9. Skeleton lines are visible during recording</td>
<td></td>
</tr>
</tbody>
</table>

A.2 Part List for Prototypes

<table>
<thead>
<tr>
<th>Item</th>
<th>Specification</th>
<th>Maker</th>
</tr>
</thead>
<tbody>
<tr>
<td>Kinect</td>
<td>RGB and depth image capture, joint coordinate</td>
<td>Microsoft</td>
</tr>
<tr>
<td>Wii Balance Board</td>
<td>Weight measure, Max 136kg, 1-DOF force transducer at four corners</td>
<td>Nintendo</td>
</tr>
<tr>
<td>Vicon MX system</td>
<td>F20-MX, 8 Cameras, IR, 2 mega pixel, Max. 500Hz</td>
<td>Vicon</td>
</tr>
<tr>
<td>Video Synchronizer</td>
<td>Trigger accuracy - &lt; 1ms</td>
<td>Kistler</td>
</tr>
<tr>
<td>OptiTrack Flex3</td>
<td>6 Cameras, 640H x 480V, IR, 100Hz</td>
<td>NaturalPoint</td>
</tr>
<tr>
<td>ATI Delta Force Sensor</td>
<td>SI-330-30</td>
<td>ATI Industrial Automation</td>
</tr>
<tr>
<td>Modification</td>
<td>Description</td>
<td>Brand</td>
</tr>
<tr>
<td>-----------------------</td>
<td>------------------------------------------------------------------------------</td>
<td>-------------------------------</td>
</tr>
<tr>
<td>ATI Nano 17 Force Sensor</td>
<td>SI-12-012, Max.: Fx,y-12N, Fz-17N, Tx,y-120Nmm, Tz-120Nm, Res.: Fx,y-1/320N, Fz-1/320N, Tx,y – 1/64Nmm, Tz-1/64N/mm</td>
<td>ATI Industrial Automation</td>
</tr>
<tr>
<td>Hexapod 6 DOF Motion Platform</td>
<td>Workspace: ±10cm(x), ±10cm(y), ±9.5cm(z), 36.81 deg (roll), 36.81 deg (pitch), 50.19 deg (yaw) Max. Speed: 0.67 m/s (x), 0.67 m/s (y), 0.35 m/s (z), 152.79 deg/s (roll), 152.79 deg/s (pitch), 80.62 deg/s Total Moving Mass: 40kg</td>
<td>N/A</td>
</tr>
<tr>
<td>Arduino Mega</td>
<td>ATmega1280, 54-Digital I/O, 16-Analog, 16MHz</td>
<td>Arudino</td>
</tr>
<tr>
<td>High Torque Servo</td>
<td>S9156, Torque - 19.6kg/cm</td>
<td>Futaba</td>
</tr>
<tr>
<td>IMU Sensor</td>
<td>VN-100S, Roll - ±180°, Pitch - ±90°, Angular Resolution - &lt; 0.05°</td>
<td>Vectornova</td>
</tr>
<tr>
<td>IMU Sensor</td>
<td>MPU6050</td>
<td>InvenSense</td>
</tr>
<tr>
<td>3D Prtinter</td>
<td>Makerbot Replicator 5th Generation Build volume: 25.2L × 19.9W × 15.0 H Layer Resolution: 100 micron</td>
<td>Makrbot</td>
</tr>
</tbody>
</table>

A.3 References for Hardware Interface (Matlab & Simulink)

Kinect Interface
http://www.mathworks.com/hardware-support/kinect-windows.html

Wii Balance Board Interface
http://netscale.cse.nd.edu/twiki/bin/view/Edu/WiiMote
http://www.colorado.edu/neuromechanics/research/wii-balance-board-project

Arduino
http://www.mathworks.com/hardware-support/index.html?c[]=hardware_catalog&q=Arduino&submitSearch=Search&s_v1=56805868_1-MB25VM
http://arduino.cc/en/Main/Software

End of the Dissertation