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FUNCTIONAL KNEE BRACE MIGRATION: BIOMECHANICAL AND NEUROMUSCULAR ALTERATIONS

by

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B.Sc., University of Ottawa, 2003

THESIS

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TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

ABBREVIATIONS AND DEFINITIONS

Angular Impulse
: The time integral of the resultant moment of force acting on a body, measured in N.m.s.

Global Coordinate System (Inertial Reference System)
: Defines the coordinate system of the laboratory from which all positions are derived. The global coordinate system is a right-handed orthogonal system (Hamill & Selbie, 2004).

FKB
: Functional knee brace

Integrated Electromyography (iEMG)
: The summation of rectified muscle activity over a period of time (Kamen, 2004).

Local Coordinate System (LCS)
: A reference system that is fixed at the centre of mass of a body or segment. The orientation of the local coordinate system to the global coordinate system defines the position and orientation of the body or segment in space (Hamill & Selbie, 2004).

Shell Brace
: Type of functional knee brace that employs molded shells of plastic or similar material connected to hinges and held in place on the leg by means of straps.

Soft Shell Brace
: Type of functional knee brace that employs elastic knitted or similar material connected to hinges and held in place on the leg by means of straps.

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ABSTRACT

Functional knee bracing has been shown to alter lower limb joint mechanics, which may protect the anterior cruciate ligament. Many knee braces have been studied, however, the effects of brace alignment or brace type on lower limb joint mechanics are not known. This study was conducted to determine whether the use of a functional knee brace, the type of brace used or its alignment relative to the knee causes biomechanical or neuromuscular alterations to gait. Ten healthy participants took part in all conditions: walking with a shell and soft shell brace, each aligned according to the manufacturers' specifications; walking with each brace distally misaligned by 1 cm as well as walking without a brace. A motion analysis and force plate system was used to determine the three-dimensional angular impulse, peak joint moments and peak joint angles of the ankle, knee and hip. Electromyographic data were used to support the kinetic data. All data were time normalized to one stride. In addition, kinetic data were normalized to body mass and kinematic data were normalized to the standing position. In comparison to unbraced walking, the shell brace in its aligned position significantly reduced the peak ankle plantarflexor moment. Additionally, there was a decreased peak knee flexion angle with the aligned shell and soft shell braces as well as an increased peak knee adduction angle and a reduced peak knee internal rotation angle with the aligned shell brace. Although there were alterations in lower limb mechanics during walking, induced primarily by the shell type functional brace, these changes were not deemed to considerably alter joint loading at the knee.

Keywords: functional knee brace, knee orthosis, anterior cruciate ligament (ACL), gait analysis

Chapter I

INTRODUCTION

Since the 1960s, there have been a large number of brace designs appearing on the market to resolve the problem of knee instability in the active population. Although some researchers believe that knee braces positively affect the performance of individuals with knee instability, there has only recently been a campaign to quantitatively study their effectiveness (Cawley, France & Paulos, 1991). Due to this apparent lack of objective data concerning knee bracing, the American Academy of Orthopaedic Surgeons (AAOS) held a symposium in August of 1984 to determine the biomechanical benefits of bracing. During this conference, three classifications of knee braces were suggested: prophylactic, rehabilitative and functional (Vailas and Pink, 1993).

Prophylactic knee braces are primarily designed to decrease the likelihood or severity of injury in healthy individuals. Rehabilitative braces allow for controlled movement of the injured operative or nonoperative knee, which is thought to offer some protection to the soft-tissue structures during the healing process. Functional knee braces (FKB) are designed to provide functional support to an unstable knee on return to activity, thereby restoring athletic performance to pre-injury levels (Arnheim and Prentice, 2000; Vailas and Pink, 1993).

Whereas subjective analysis has previously been the primary mode of functional brace evaluation, it has only been in the past two decades that quantitative biomechanical analyses have been performed. As a result, analysis of the published research does not yield one unified conclusion. The disagreement between findings may be due, in part, to differences in research design and instrumentation among the studies (Vailas and Pink, 1993).

Among the published subjective analyses, however, one common complaint of participants is distal brace migration–the tendency of the brace to gradually slide down the leg during activity (France, Cawley & Paulos, 1990; Greene, Hamson, Bay & Bryce, 2000). Regalbuto, Rovick and Walker (1989) have proposed that brace migration may be due to the failure of the brace to adhere to the soft tissues of the leg during the successive contractions that are characteristic of physical activity. As the main function of the brace is to accept some of the applied force during activity while restricting anterior tibial translation, the axes of rotation of the knee and the brace must remain aligned during activity to maintain maximum protective benefit. To date, there have been few studies that have attempted to quantify functional knee brace migration (Brownstein, 1998; Miller, Vailas & Croce, 1999; Greene et al., 2000; Rast, 2000; Wojtys & Huston 2001; Lamontagne, Singer & Xu, 2003). Nevertheless, these studies have found that brace migration does occur, which reduces the comfort and possibly the efficacy of the brace and may lead to decreased user compliance and mechanical benefit.

Brace migration is believed to occur irrespective of the brace design. Strap braces migrate considerably more than shell braces (Rast, 2000), although there is uncertainty regarding the superiority of custom-fit braces over off-the-shelf varieties (Vailas & Pink, 1993; Brownstein, 1998; Wojtys & Huston, 2000). Moreover, it has been determined that an offset of more than 12 mm between the knee and brace axes of rotation will lead to increased forces and moments within the hinge, which may lead to shearing of the soft tissues of the leg, altered joint mechanics, reduced ability of the brace to control anterior tibial translation and, consequently, increased anterior cruciate ligament (ACL) strain (Regalbuto et al., 1989).

Statement of the Problem

There has been extensive research on knee joint moments and ACL strain during locomotion (DeVita, Torry, Glover, & Speroni, 1996; DeVita, Lassiter, Hortobagyi, & Torry, 1998; Shelburne, Pandy, Anderson, & Torry, 2004b; Shelburne, Pandy, & Torry, 2004a). From this research, the knee extensor moment caused by the anteriorly directed force of the rectus femoris on the tibial tuberosity is related to strain and force on the ACL during walking (DeVita et al., 1998). Pandy and Shelburne (1997) developed a sagittal plane model to estimate the peak ACL force during walking, which was found to reach approximately 300 N at contralateral toe-off. It has been documented that functional bracing will cause changes in knee mechanics that reduce the strain and force in the ACL (DeVita et al., 1996; DeVita et al., 1998). More recently, it has been proposed that elevated knee moments in the frontal and transverse planes may be factors that play a role in the development of patellofemoral pain syndrome in runners (Stefanyshyn, Stergiou, Lun, Meeuwisse & Nigg, 1999). Moreover, there is a relationship between peak external knee adduction moments and knee osteoarthritis (Baliunas, Hurwitz, Ryals, Karrar, Case, Block & Andriacchi, 2002), though it is unclear as to whether this factor is causative of osteoarthritis.

While knee bracing will cause changes in gait mechanics of healthy individuals (DeVita et al., 1996), it is unclear as to whether brace migration leads to biomechanical or neuromuscular alterations that at the knee may modify ACL strain or joint loading.

With this being the case, the purpose of this investigation is threefold: (1) to determine if shell or soft-shell type functional knee braces alter lower limb joint mechanics in comparison to non-braced walking; (2) to determine if misalignment of each brace alters joint lower limb joint mechanics and (3) to determine if the effects of hinge misalignment are dependent on the brace type. The goal of this research is to shed light on both the efficacy and safety of functional knee bracing. It is hoped that this will ultimately provide information to athletes and medical practitioners so that informed decisions can be made regarding the advantages and drawbacks of functional knee brace use.

Hypotheses

The following hypotheses were initially proposed based on previous research (Branch et al., 1988; Walker et al., 1988; Regalbuto et al., 1989; Vailas & Pink, 1993; DeVita et al., 1996; DeVita 1998; Rast, 2000).

- Increase in ankle plantarflexion angular impulse and hip extensor angular impulse in comparison to normal unbraced walking – largest increase for the shell brace, followed by the soft shell brace.
- Increase in peak moment values for the ankle and hip, with differences being largest for the shell brace and soft shell brace, respectively.
- Reduced knee flexion, peak extensor moment and angular impulse in comparison to unbraced walking due to the increased rigidity that the brace imparts to the knee – greatest difference for the shell brace in comparison to unbraced walking.
- Increase in hip abductor angular impulse and peak moment due to the reduced knee flexion and, hence, the increased need of the hip abductors to clear the limb during swing –greatest in the shell brace followed by the soft shell brace in comparison to unbraced walking.
- The abovementioned differences should also persist when comparing aligned and misaligned conditions, but should be exacerbated by misalignment of the brace.

Limitations

The proposed investigation has inherent limitations. Firstly, the use of skin-mounted markers to measure skeletal kinematics has been shown to be less accurate than intracortical pin-mounted markers (Reinschmidt, van Den Bogert, Nigg, Lundberg & Murphy, 1997; Ramsey and Wretenberg 1999; Ramsey, Lamontagne, Wretenberg, Valentin, Engstrom & Nemeth, 2001; Benoit, Ramsey, Lamontagne, Wretenberg, & Renstrom, 2005), especially in the frontal and transverse planes. The error in kinematic data derived from skin-mounted markers is assumed to be relatively consistent across subjects and will be controlled for using the repeated measures design. Secondly, the use of the soft shell brace may introduce noise into the EMG signal, as it will be applying pressure to the electrodes on the vastus medialis and gastrocnemius. Proper placement and testing of the electrodes prior to the trials should at least minimize the associated error. Furthermore, the use of healthy participants within the study may limit its application to ACL-injured patients but is considered a necessary compromise, such that differences in knee joint moments and kinematics between the conditions can be attributed to the brace itself, rather than the effect of differences in the degree and nature of injury between participants. Lastly, the applicability of the study can only extend to shell or soft shell braces that employed the same hinge design as those used in the study.

CHAPTER II

REVIEW OF LITERATURE

The knee is one of the most complex joints in the human body. Because the knee is positioned between the body's two longest lever arms – femur and tibia – it is subject to very high forces and, hence, is vulnerable to injury (Wirth and DeLee, 1990; Nordin and Frankel, 2001). These anatomical considerations, coupled with the recent rise in participation in athletic activities due to health promotion and the fitness boom, have caused an increase in various forms of injury to the knee – more specifically to the anterior cruciate ligament (ACL). As a result, the use of functional knee braces (FKBs) has become the standard convention for restoring normal motion to the nonoperative ACL deficient knee and reducing the strain on the postoperative ACL graft.

Over two decades have passed since the American Academy of Orthopaedic Surgeons held its symposium on knee braces. There is still, however, much disagreement with respect to the effectiveness of functional knee braces in reducing the strain on the ACL graft or reducing anterior translation of the tibia in the ACL deficient knee (Vailas and Pink, 1993; Wojtys, Kothari & Huston, 1996; Beynnon and Fleming, 1998; DeVita et al., 1998; Fleming, Renstrom, Beynnon, Engstrom & Peura, 2000; Ramsey et al., 2001). The following section will review various static and dynamic analyses that endeavoured to determine the effectiveness of functional knee braces to protect the ACL from strain. Hinge design and placement will also be discussed with regard to its effect on ACL strain.

Functional ACL Deficiency and Knee Brace Use

Functional Assessment and Quasi-Static Analysis

Shortly after the symposium on knee braces was held, there was a rise in the number of biomechanical investigations of functional knee braces. Most of these studies employed functional assessment tests such as the anterior drawer, Lachman drawer, and the pivot-shift test to examine the benefit of using these orthoses. Colville, Lee and Cuillo (1986) examined anterior tibial translation in forty-seven ACL deficient knees using the Lachman test and found that absolute anterior tibial translation was unaffected by the Lenox Hill brace, although the resistance to anterior translation was increased – indicating that the brace had increased the stiffness of the joint. The authors also demonstrated that axial rotation of the tibia was clinically reduced in 80% of the participants. This finding signified a potential to reduce ACL strain or injury, as tibial axial rotation has been implicated in the mechanism of non-contact ACL injury. It should be noted that, in a similar study with the same brace, Bassett & Fleming (1983) found that only 50% of the patients had improved rotatory control with the brace.

Due to these discrepancies, these studies have been criticized on the basis of reliability and validity because of the subjectivity of clinical assessments. Moreover, the terminal segment was not maintained in a fixed position and was not axially loaded, which is atypical of the usual non-contact mechanism of ACL injury (Vailas and Pink, 1993). Axial loading of the tibia has been thought to increase the stiffness of the knee joint, compressing the articular surfaces of the femur and tibia. Axial loading is also believed to increase the congruence of the articular surfaces, which increases their contact surface area and, hence, increases tibial resistance to anterior translation (Walla, Albright, McAuley, Martin, Eldridge & El-Khoury, 1985; Wojtys and Huston, 1994; Wojtys et al., 1996; Liu and Maitland, 2000). Tests performed without axial loading of the tibia were not representative of in-vivo tibial translation. Interestingly, despite its potential stabilizing effect, axial loading of the tibia might cause anterior tibial translation if there is corresponding contraction of the quadriceps. Anterior displacement of the tibia would be due to the anteriorly directed force that the quadriceps exerts via the patella onto the tibial tuberosity.

In remedy to the limitations of these studies, investigators studied the effects of the functional knee brace on its ability to resist anterior tibial translation with applied axial loading of the tibia during active contraction of the leg musculature. Wojtys et al. (1996) performed such a study, which demonstrated that during “muscles-active” trials, there were significant reductions in anterior tibial translation with the use of a functional brace with respect to “muscles-relaxed” trials. The authors noted that, in comparison to the unbraced condition, the brace was able to reduce anterior tibial translation by 33.1% when the leg musculature was relaxed and 80.1% when the musculature was contracted. The authors also indicated that five of the six braces that were tested reduced hamstring torque, average work, average power and work fatigue with no corresponding decrease in quadriceps muscle performance. Although both the quadriceps and hamstring muscle groups served to compress the knee joint and exert opposing forces on the tibia, a deficiency in hamstring muscle performance may have been detrimental to the joint. Without adequate hamstring performance, the quadriceps may be capable of causing anterior translation of the tibia relative to the femur through its action on the tibial tuberosity. Furthermore, in their study, Wojtys et al. (1996) determined that quadriceps and hamstring reaction times slowed with knee brace use. The authors noted that with slowed hamstring muscle reaction time and performance during bracing, the leg might be more vulnerable to anterior tibial translation and subsequent injury to the ACL – especially if neither the performance nor reaction time of the quadriceps is reduced. This has also caused concern, as the lag in muscle response time seen with functional knee braces is in the same order as the time sequence from force onset to ACL failure in the in-vivo cadaveric model (Wojtys et al., 1996). As an alternative to the idea that functional knee bracing hampers muscle performance and reaction time, Branch, Hunter & Donath (1989) proposed that these effects may be due, in part, to an adaptive effect of bracing, whereby the brace stabilizes the knee and reduces the need for muscle activity.

Muscle Performance

Results from studies on muscle performance with the use of knee bracing have been equivocal. In a two-year longitudinal study on prophylactic knee bracing in college football, Rovere, Haupt and Yates (1987) found that rates of knee injury were higher when the braces were worn, as compared to a similar period when they were not. The authors indicated that the AAOS believed that knee stabilizing devices may preload ligamentous structures in certain positions, and this may be the reason for the increased injury rates observed. In terms brace comfort, players complained of lower limb muscle cramping when wearing the brace – most likely due to the tensile force in the straps, leading to the application of pressure on the leg musculature. Furthermore, Houston and Goemans (1982), in a study on leg muscle performance both with and without a knee brace, determined that maximal isometric torque outputs during knee extension were significantly lower by 12-30% in the braced condition, with the differences being accentuated during faster contractions. Maximal unloaded knee extension velocity was also reduced by the brace and blood lactate concentrations during 15-minutes of fixed-load cycling on an ergometer elicited a 41% higher concentration in the braced condition. They speculated that braces could interfere with blood flow and oxygen delivery to muscles. In a similar study, Birmingham, Kramer and Kirkley (2002) found that this reduction of extensor torque was dependant on patient strength, with detriments being greater for stronger patients. The authors also stated that, for weaker patients, the brace yielded no significant change and could possibly even yield a benefit in terms of strength.

These particular findings are in parallel with the increased cramping found by Rovere et al. (1997) and with studies by Osternig and Robertson (1993) and Smith, Malanga, Yu and An (2003). These authors sought to determine the changes in muscle activation patterns that occur with knee bracing. Osternig and Robertson (1993) showed that the knee flexors, extensors, gastrocnemius and tibialis anterior were highly sensitive to bracing. Specifically, bracing reduced EMG activity in 73% of the significant braced versus nonbraced comparisons. The previous results indicated that any mechanical benefit of the brace, in terms of stabilizing the inert structures of the knee, come at the expense of muscle performance. Smith et al. (2003), examined onset times of the muscles that cross the knee joint during a single leg hop manoeuvre. A favourable contraction pattern was defined as an earlier reflex firing of the hamstrings relative to the quadriceps. The authors found that knee bracing did not consistently result in this more favourable muscle firing pattern. Whereas five of ten subjects contracted the hamstrings or gastrocnemius first prior to impact without the brace, only an additional two used the more favourable muscle firing pattern with the brace.

Studies indicating increased cramping and blood lactate concentration as well as altered muscle activation patterns may at least be partially explained by studies of intramuscular pressure during bracing. Although the application of a brace using a high strap tensile force may resolve the problem of hinge-to-knee center of rotation mismatch, problems may ensue due to reduced blood flow to the lower limb. It has been shown that external compression of the limb from brace use causes a significant reduction in muscle blood perfusion during exercise within the anterior compartment of the leg (Styf, 1990; Styf, Nakhostine & Gershuni, 1992; Styf, Lundin & Gershuni, 1994; Lundin and Styf, 1998) leading to a reduced time to fatigue of these muscles (Sforzo, Chen & Gold, 1989; Styf et al., 1994). Moreover, Styf et al. (1992) observed that removing the distal strap from the brace returned intramuscular pressures in the anterior compartment to near normal values, while Beynnon, Johnson, Fleming, Peura, Renstrom, Nichols and Pope (1997) reported that lowering the distal strap tension from 45 N to 22 N did not reduce the protective effect of the brace on the ACL. It should be noted, however, that intramuscular pressures are increased to values that will significantly reduce time to fatigue even at strap tensile forces that are deemed comfortable to the participant (Lundin and Styf, 1998). Similar effects of functional bracing on muscle performance in the rectus femoris were found (Lundin and Styf, 1998), and it has been thought that bracing has similar effects on virtually all leg muscles that cross the knee joint.

ACL Strain

Results from studies of the effectiveness of FKBs to reduce ACL strain have also resulted in ambiguous conclusions. Initial direct ACL strain measurements with implanted strain gauges in cadaver limbs led to the conclusion that functional knee braces increased ACL strain with passive flexion-extension of the knee joint (Beynnon & Fleming, 1998). In this study there appeared to be a pre-loading of the ACL, which suggested that FKBs may not be warranted in ACL deficient subjects and could even be harmful. Follow-up studies of ACL strain in human subjects by Beynnon, Pope, Wertheimer, Johnson, Fleming, Nichols and Howe (1992) revealed that there was in fact a strain-shielding effect caused by both a custom-fit and an off-the-shelf type of functional knee brace during static conditions. This effect was deemed to be dependent on the magnitude of the anterior shear load applied to the tibia, as the ability of the brace to reduce strain in the ACL diminished at 140 N of applied load. The authors also noted that neither brace type had any effect on ACL strain during active motion of the knee (10 to 120 degrees) or during isometric contraction of the quadriceps. In contrast to earlier studies, Beynnon et al. (1992) found that bracing did not increase ACL strain values and both types of brace were equally effective at low magnitudes of load.

A further study by Beynnon et al. (1997) sought to determine the effect of weightbearing and functional bracing on ACL strain. Using an implanted differential variable reluctance transducer (DVRT) in the ACL, the authors found significantly more ACL strain when weightbearing as opposed to non-weightbearing, while similar strain values were noted with the application of an anteriorly directed shear load (140 N) to the tibia. This indicates that the ACL is strained during weightbearing and the compressive load on the knee joint caused by body weight and active contraction of the knee musculature does not reduce the strain on the ACL with an applied shear load. The use of a functional brace, however, did provide protection to the ACL during both weightbearing and non-weightbearing, with and without applied shear loads. Fleming et al. (2000) also found similar results during weightbearing versus nonweightbearing trials. In the nonweightbearing condition, the overall effect of bracing on ACL strain was dependent on the magnitude of the applied shear force, with the strain shielding effect of the brace levelling off at about 130 N. In the weightbearing condition, however, ACL strain was not dependent on the magnitude of the applied force. These findings are consistent with another investigation by Beynnon and Fleming (1998) who found that, during weightbearing, the ability of the brace to reduce anterior tibial translation remained constant with an increasing magnitude of anterior force.

Furthermore, it was determined that internal axial rotation significantly increased ACL strain in the unbraced condition and bracing significantly reduced this strain. Again, the strain shielding effect of the brace was dependent on the magnitude of the load – as internal axial rotation approached 9 N·m, the brace failed to significantly reduce ACL strain.

One limitation of the aforementioned studies is that the applied loads in the anterior direction were limited to relatively low values (< 200 N). As a result, we cannot extrapolate the results and speculate on anterior tibial translation and ACL strain during physiologic loading (Vailas and Pink, 1993; Fleming et al., 2000), during which anterior shear force applied to the tibia is thought to exceed 250 N (Taylor, Walker, Perry, Cannon & Woledge, 1998; Shelburne et. al., 2004a).

Dynamic Analysis

Another method of evaluating functional bracing is through dynamic biomechanical analysis. This method allows researchers to examine the biomechanics of the limb in braced and non-braced conditions during various forms of exercise. DeVita, Blakenship-Hunter and Skelly (1992) investigated the biomechanics of running with a knee brace and failed to show any differences in kinematics between braced and non-braced conditions among ACL deficient participants. However, due to the fact that bracing caused a significant increase in hip extensor torque and a decrease in knee extensor torque, the authors indicated that the ACL patients may have adapted to the effects of the brace during the rehabilitation period, and hence also ran without the brace using these augmented movement patterns. Furthermore, kinematic analysis revealed that the ACL patients ran with a more erect body position throughout the stance phase. This finding has been cause for concern, as the decreased flexion angles at the knee and hip, in addition to the more stiff running style, may provide lesser amounts of shock absorption. As a result, the ACL deficient patient, in attempting to reduce the amount of anterior tibial displacement, may be paving the way for future meniscal injuries.

Follow-up studies by DeVita et al. (1996) and DeVita et al. (1998) revealed significant differences in knee flexion during the stance phase in the braced condition compared to the unbraced condition. Moreover, it was found that the use of a brace increased the extensor angular impulse and moment at both the hip and ankle, while knee angular impulse and extensor moment were significantly decreased. This has been shown to exert a protective effect on the knee, as loads on the ACL are directly related to the extensor moment at the knee. The authors were aware that previous studies had failed to identify a protective effect of functional bracing on the knee during loading at physiologic levels. As a result, the authors attempted to quantify the neuromuscular and motor pattern adaptations that occur with functional bracing, rather than merely assessing the direct mechanical effect of the brace on the knee joint (DeVita et al., 1998). They concluded that, while the use of functional braces is not a necessity in the ACL deficient patient or the postoperative patient, braces do promote neuromuscular and motor pattern adaptations that are conducive to ACL protection. Campbell, Armstrong and Cipriani (2002) also assessed the alterations in gait pattern seen with functional knee bracing. In line with the results of DeVita et al. (1996), the authors determined that functional knee bracing does promote adaptations in gait pattern that may be beneficial in protecting the ACL. Furthermore, as the changes in gait pattern persisted across bouts of exercise and over a twenty-four hour period of time, the authors believed that the effect caused by functional knee bracing may persist over time.

In agreement with the studies by DeVita et al. (1992) and DeVita et al. (1998), Shelburne et al. (2004a), with the use of a two-model simulation approach using both forward and inverse dynamics, found that during the stance phase of walking the total anterior shear force applied to the tibia was less in ACL deficient subjects than in healthy subjects, with values of 76 and 259 N, respectively. The authors found that the peak anterior shear force occurred at contralateral toe-off for both ACL deficient and healthy subjects while there were no differences between groups in shear knee force during swing.

DeVita and Hortobagyi (2001) used a similar mathematical approach and also found a reduction in peak anterior knee shear force in ACL deficient subjects versus normals. Compared to healthy individuals, ACL subjects walked with greater hamstrings impulse and less quadriceps impulse. Further to this, functional knee bracing increased the hamstrings impulse and decreased the quadriceps impulse in the ACL subjects, with corresponding decreases in anterior shear knee force in both the ACL deficient and normal subjects. Rather than identifying neuromuscular adaptations as the cause for the altered knee mechanics observed in ACL deficient individuals, Shelburne et al. (2004a) indicated that a change in the patellar tendon angle, caused by anterior tibial translation, may reduce the total anterior shear force at the knee. These results indicated that the adaptation that occurs after ACL injury may not be simply an alteration in the motor pattern of gait, but could also be caused by changes in the anatomy and mechanics of the knee following an ACL rupture.

Hinge Design and Placement

The knee brace attempts to displace the loading on the knee by constraining its motion and bearing some of the force imposed during activity. With this being the case, it is crucial that the kinematics of the brace match, as closely as possible, the kinematics of the normal knee; more specifically, the axis of rotation of both the knee and the brace should be congruent. Currently, braces have a fixed or polycentric axis, though the polycentric axis has been deemed to more closely match the kinematics of the normal knee. Active and passive brace designs have also been proposed, whereby the active design applies a posterior force to the tibia to limit anterior translation.

Brownstein (1998) measured the migration tendencies of 14 commonly used functional knee braces and found that each brace migrated a measurable amount. The extent of brace migration was not influenced by brace type (custom or off the shelf) or fit method (cast or measuring tool) but was limited in braces that employed an active design. These results are somewhat contrary to those of Wojtys and Huston (2001), who found that brace migration was significantly greater in a custom brace (CE-2000) than for an off the shelf brace (Goldpoint). Whereas Wojyts and Huston (2001) had distal migration values of 18.6 mm for the CE-2000 brace and 4.5 mm for the Goldpoint brace, Brownstein (1998) reported values of 3 mm and 4.38 mm, respectively. Differences between reported distal migration values may be due to differences in study protocol. Wojtys and Huston (2001) had participants use a bicycle or stairmaster for 20 minutes followed by an additional 40 minutes of any activity; Brownstein (1998) had participants use a commercial stair-climber, treadmill and a slide board, each for 15 minutes. From this, the increased amount of migration reported by Wojtys and Huston (2001) may be due to the length of time participants exercised, the large knee range of motion during cycling or from the potential for higher impact activity that may have been part of the 40 minute exercise bout.

A study by Regalbuto et al. (1989) investigated the effect of hinge design and placement on the forces acting on the hinge itself. A mismatch between the axis of rotation of the knee and the brace was defined in terms of the forces and moments at the hinge. Their findings indicated that higher forces and moments at the hinge occurred during flexion of more than 60° and when the hinge was offset more than 12 mm from the ideal location, irrespective of the type of hinge studied. This mismatch between the brace and the knee has been thought to translate into a reduced ability of the brace to control tibial motion, which could further lead to increased ACL tension (Walker et al., 1988). Additionally, mismatch may cause discomfort due to the shearing of the soft tissues beneath the brace, as well as pistoning and migration (Regalbuto et al., 1989; Vailas and Pink, 1993).

A study by Lamontagne, Singer and Xu (2003) sought to determine the extent of relative motion that occurs between the leg and a soft shell brace during cycling and establish how this motion changes over time. The authors determined that while brace migration did not occur in its definitive sense, there was a constant disparity between the kinematics of the knee and the brace, which persisted throughout the 15-minute testing session. Differences between the motion of the leg and brace, measured in the sagittal plane, were most likely the result of slight movement of the brace along the longitudinal axis of the leg due to bunching of the fabric of the brace behind the knee. Dissimilarities in the motion of abduction/adduction were perhaps caused by the rigidity of the lateral supports of the brace and, hence, their ability to resist bending along their length. The leg, however, was free to abduct and adduct about the knee within the limits afforded by the elastic material of the brace. Lastly, it was thought that disparity between the kinematics of the leg and brace in terms of internal/external rotation was caused by motion of the leg about its longitudinal axis within the brace. The authors noted that the disparities in kinematic patterns within the three planes of motion were largest when maximum flexion occurred–this is in keeping with the study by Regalbuto et al. (1989).

Due to these possible detrimental effects of hinge-to-knee mismatch, custom-made braces, which have been shown to better correspond with the contours of the leg, may remedy this problem (Walker et al., 1988; Regalbuto et al., 1989; Cawley et al., 1991; Vailas and Pink, 1993; Rast, 2000). Alternatively, off-the-shelf shell or strap braces, which are sized according to population parameters, may also consistently match knee kinematics if they closely approximate the joint anthropometrics and are applied tightly (Walker et al., 1988; Vailas and Pink, 1993).

Summary

Despite the plethora of scientific information surrounding the use of functional braces during exercise, there continues to be much controversy regarding its effectiveness. Some investigators claim that the risks from bracing such as slowed reaction time and decreased muscle performance due to decreased blood perfusion clearly outweigh any benefits. Conversely, others indicate that increased extensor moments at the hip and ankle and reduced moments at the knee clearly reveal the adaptive effects of bracing and its protective effect on the ACL. One factor that is particularly intriguing within the scope of this particular investigation is the relationship between strap tensile force, blood perfusion, and brace migration. It has been shown that increasing strap tensile force will allow for better congruence between the hinge and knee axes of rotation. Increasing strap tensile force, however, may decrease blood perfusion to the leg musculature, leading to shorter time to fatigue. If brace migration can either be identified or eliminated as a cause of changes in joint moments and muscle activation patterns that occur during bracing, steps can be taken to correct problems of hinge to knee mismatch as well as decreased muscle perfusion due to bracing.

CHAPTER III

METHODOLOGY

Participants

Ten healthy male participants, having neither anatomical nor gait abnormality or history of knee injury, were recruited from the student population at the University of Ottawa. Uninjured participants were used to ensure that any changes in joint kinetics and kinematics seen during analysis could be directly attributed to a change in the location of the brace axis of rotation relative to the knee. If ACL-deficient subjects were used, a confounding factor, such as subject variability in injury or the effects of fatigue on the ability of the leg musculature to maintain knee stability could be said to be causing the alteration.

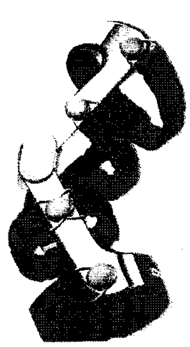
Possible anatomical and gait abnormalities were screened for by means of a gait analysis using the unbraced experimental condition. Any anatomical or gait abnormality disqualified participants from inclusion in the study. All subjects signed an informed consent form in accordance with the Research Ethics Board at the University of Ottawa.

Knee Brace Application

The DonJoy 4Ttude (shell type) and the Bauerfeind SofTec Genu (soft shell type) knee braces were applied to the participants in accordance with the manufacturers' specifications. In determining the correct brace size, measurements of the calf and thigh were taken. For the DonJoy 4Ttude brace, thigh diameter measurements were taken 15 cm superior to the mid-patella. For the Bauerfeind Softec Genu brace, thigh diameter measurements were taken 16.5 cm superior to the mid-patella; calf diameter measurements were taken 15 cm inferior to the mid-patella.

Brace application for the DonJoy 4Ttude followed the correct strap sequence. The most proximal strap was applied first, to help position the brace on the leg. The centre of the hinge joint on the brace was aligned with the superior aspect of the patella, while residing slightly posterior to the midline of the leg. Following the application of the first strap, the most distal strap was secured, followed by the second most proximal. Lastly, the posterior second most distal strap was applied, followed by its corresponding anterior strap.

The Bauerfeind SofTec Genu brace was applied such that the patella was located within the patellar support. Following this procedure the lateral supports were removed, the pivoting cam in the hinge was unlocked, the lateral supports were reintegrated into the brace and the knee was flexed and extended a number of times to align the hinge axis of rotation with that of the knee. After the axes were aligned, the cams were locked to maintain the axis of rotation of the hinge. All brace applications were performed by the principal investigator, during which the brace straps were tightened and fastened according to the subjective comfort level of each participant.



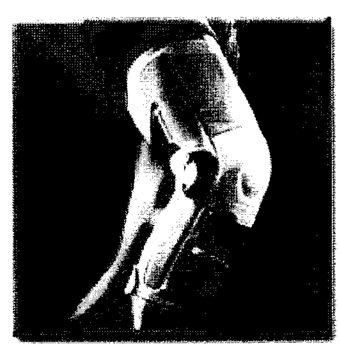


Figure 3.1 Donjoy (shell) brace (left); Bauerfeind (soft shell) brace (right)

Instrumentation

Four charge coupled device (CCD) cameras (Bassler Vision Technologies, GmbH), sampling at 80 Hz, were mounted on tripods and used to record the participants' walking trials. Cameras were positioned to cover the entire measurement area, which was determined by a calibration frame (2 x 1 x 1 metres; 30 control points). The optical axis between two adjacent cameras was at least equal to or greater than 60° (Figure 3.2). Each camera view was then verified to ensure that all calibration frame markers could be seen within the field-of-view.

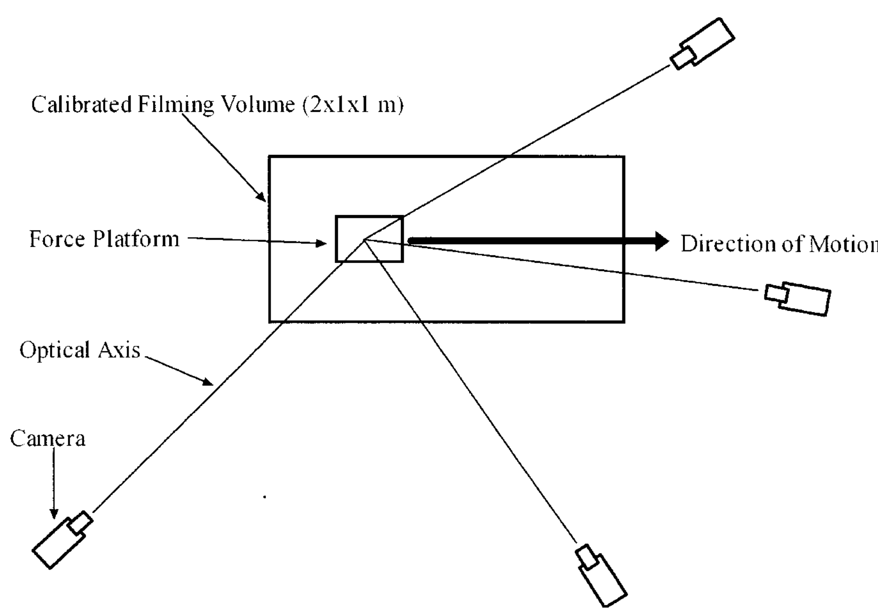


Figure 3.2 Laboratory Set-up (not to scale)

One Kistler force platform was used to measure vertical, horizontal and mediolateral ground-reaction forces at 640 Hz for the stance phase of one stride. In addition, the electromyographic signals were collected for two strides from the leg of interest. EMG electrodes were placed on the vastus medialis and lateralis, semitendinosis and biceps femoris, as well as on the medial and lateral heads of the gastrocnemius. EMG channels were also sampled at 640 Hz. Video data and all analog signals were synchronously recorded by SIMI Motion software (SIMI Motion Systems, GmbH).

Marker System and Local Coordinate System Definitions

To accurately define a local coordinate system (LCS) for the segments of the lower limb and pelvis, retroreflective calibration markers, of 1 cm diameter, were placed on the subject over anatomically relevant locations in a similar fashion to the marker set described by the National Institute of Health (Hamill and Selbie, 2004) (Figure 3.3; Table 3.1). The first step in defining the LCS was to define the plane that runs concurrently with the long axis of the segment (z-axis). This was accomplished by placing markers on medial and lateral landmarks at the proximal and distal ends of each segment. The origin of the LCS, or joint centre, was located at the midpoint between the proximal medial and lateral markers of the segment that lies distal to the joint. The vector that joins the joint centres that lie at the proximal and distal ends of the segment defined the orientation of the z-axis. The orientation of the mediolateral axis (x-axis) was defined by the vector that joins the medial and lateral markers that were used to define the location of the joint centre. The orientation of the anteroposterior axis (y-axis) was defined as the cross product of the z- and x-axes. Positive directions for the segment coordinate system axes were defined by the right hand rule: lateral for the x-axis; anterior for the y-axis; up for the z-axis (Figure 3.4). Corresponding clinical rotations were flexion-extension about the x-axis, adduction-abduction about the y-axis and internal-external rotation about the z-axis. Positive rotations about these axes were determined with the right hand rule.

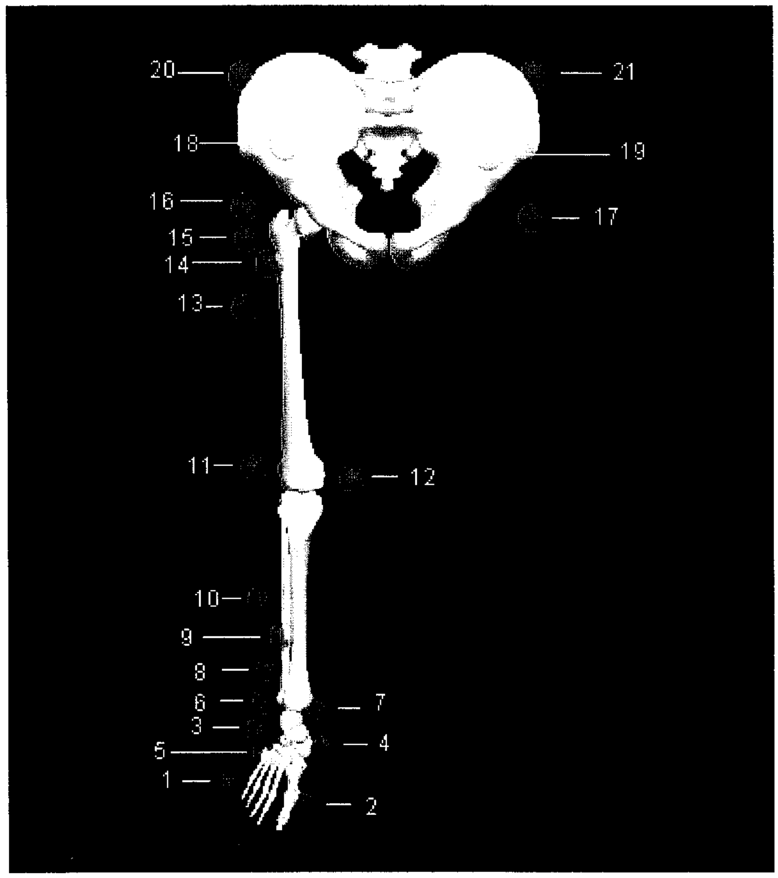


Figure 3.3 Schematic of calibration (LCS) and triad markers.

Table 3.1 Anatomical locations of calibration (LCS) and triad markers.

Marker Number	Location	Marker Number	Location
1	*Head of 5th metatarsal**	12	Medial condyle
2	Head of 1st metatarsal	13	*Distal thigh*
3	*Calcaneus lateral**	14	*Anterior thigh*
4	Calcaneus medial	15	*Proximal thigh*
5	*Forefoot*	16	Greater trochanter (right)
6	Lateral malleolus	17	Greater trochanter (left)
7	Medial malleolus	18	*ASIS (right)**
8	*Distal shank*	19	*ASIS (left)**
9	*Anterior shank*	20	*Iliac crest (right)*
10	*Proximal shank*	21	*Iliac crest (left)*
11	Lateral condyle		

Note: Triad markers are listed in *italics*, while markers that serve as both triad and calibration markers are labelled with an asterisk.

Exceptions to the abovementioned conventions were made for the calculation of the hip joint centre, which was located medial to the greater trochanter at 11% of the distance between the right and left ASIS (Bell, Brand & Pedersen, 1989). In addition, the orientations of the LCS axes for the foot were altered, as the long axis is concurrent with the plane that defines its dorsal aspect. From this, the positive x-axis is directed laterally, the positive y-axis is directed up and the positive z-axis is directed posteriorly (Figure 3.4). Corresponding clinical rotations were plantarflexion-dorsiflexion about the x-axis, internal-external rotation about the y-axis and inversion-eversion about the z-axis.

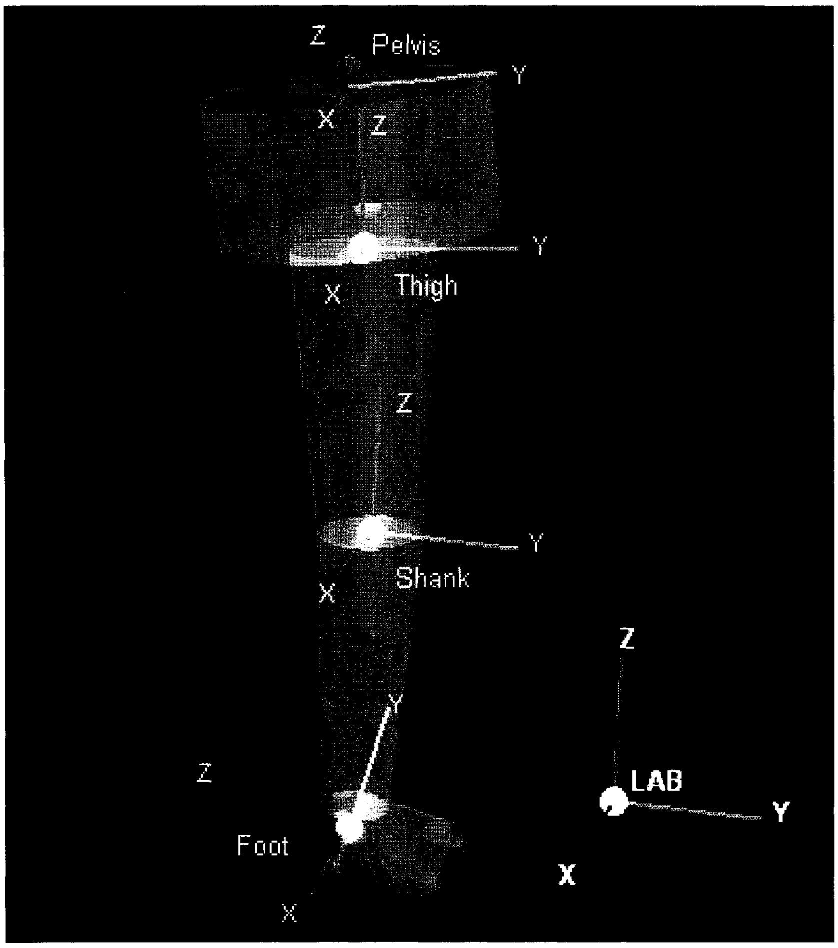


Figure 3.4 Representation of LCS axes for ankle, knee and hip with respect to the laboratory coordinate system

Three non-collinear triads of markers, placed on the foot, leg and thigh were used to determine the 3D kinematics of each respective segment. An additional four markers, placed on the right and left iliac crest and ASIS, were used to define the position and orientation of the pelvis.

Experimental Set-up and Protocol

The force platform was located at the origin of the inertial reference system, or global coordinate system. Axes for the global coordinate system were oriented in a similar fashion to the segment axes of the knee, hip and pelvis, where the positive x-axis was directed horizontally and perpendicular to the axis of motion, the positive y-axis was coincident with the plane of progression and the positive z-axis was directed up (Figure 3.4).

Prior to data collection, participants were asked to perform a standing reference trial, where they were to stand at the origin of the global coordinate system and align themselves, as best as possible, with this axis system. The standing reference trial was used in determining the transformation matrix describing the position and orientation of the local coordinate system in the global coordinate system as well as the position of the triad markers on the limb.

Electrodes were placed on the aforementioned muscles of interest after shaving and cleansing of the underlying skin to reduce skin resistance. A maximal voluntary contraction of each muscle of interest was used to set the gain for each channel.

Participants were randomly assigned to take part in two walking conditions for each of the two braces – an aligned condition, where the brace was positioned on the leg according to the manufacturers' specifications and a misaligned condition where the axis of rotation of the brace had been deliberately shifted distally by 1 cm. An unbraced condition was also performed to serve as a baseline measure. Five trials of data were captured for each of the five experimental conditions. The relative position of the brace on the leg during both aligned and misaligned conditions was determined dynamically during each trial by placing markers on medial and lateral hinge centres. Medial and lateral hinge marker trajectories in reference to the LCS axes of the shank were compared to the position of the knee joint centre, as determined with the standing reference trial.

Participants were asked to walk at self-selected speeds, impacting the force platform with the instrumented leg on the second stride. Videographic data were captured for one stride, beginning at heel-strike on the force platform and ending at subsequent heel-strike. Electromyographic data were collected for two successive strides, beginning at heel strike on the force platform, while force platform data were recorded for the duration of stance phase. Each participant was tested during a single session using his own running shoes.

Data Reduction and Statistical Analysis

All standing and measurement trials were digitized with the SIMI system. Two-dimensional image coordinates from each camera were transformed into three-dimensional marker position data using SIMI's Direct Linear Transform algorithm.

Three-dimensional marker coordinates, calculated force platform and raw EMG data from each trial were exported separately in *.TXT* format. Marker and force data files were manually trimmed and time-synchronized to heel-strike, imported into Bioproc2 (Robertson, 2005) and converted to *.FSV* format for import into Visual 3D (C-Motion Systems, Inc.). Using Visual 3D software, 3-D marker data were filtered using a zero-lag, twentieth-order, Critically Damped, low-pass filter with a cut-off frequency of 4.0 Hz (Robertson &

Dowling, 2003). Force platform data were filtered with a zero-lag, fourth-order, Butterworth, low-pass filter with a cut-off frequency of 15 Hz.

The lower extremity was modeled as a rigid, floating-segment system. Foot, shank and thigh segments were modelled as frustra of right cones, while the pelvis was modelled as a right elliptical cylinder. Segmental mass, moment of inertia and centre of gravity location were determined by Dempster's anthropometric data, using the segment length, defined by the proximal and distal segment ends and the segment geometry. A standard Newton-Euler method was used to calculate the joint reaction forces and internal moments at each joint. Joint moments were resolved into the LCS of the distal segment and were normalized to body mass.

Joint angular displacement data were calculated by the rotation matrix describing the transformation of the distal LCS to the proximal LCS, resolved into the LCS of the distal segment. Joint angular displacement data were described using the Cardan notation and were normalized to the standing position. As the calculation of Cardan angles is sequence dependent, the Xyz order was chosen for the hip and knee. For the ankle, the Xzy sequence was chosen, such that the ordered sequence about anatomical axes would be the same for all three joints. From this, the ordered sequence corresponds to rotations about the flexion-extension (or plantar-dorsiflexion) axis, abduction-adduction (or inversion-eversion) axis and internal-external rotation axis. This ordered sequence of rotations about anatomical axes, while about differently labelled LCS axes, has been recommended as a standard in biomechanics (Cole, Nigg, Ronsky & Yeadon, 1993).

Electromyographic data from each stride were filtered in Bioproc2 (Robertson, 2005) with a twentieth-order, Critically Damped, high-pass filter, with a cut-off frequency of 20 Hz

(Robertson & Dowling, 2003). Data were then full-wave rectified and time integrated, yielding the total muscular activity for the duration of stride.

Peak positive and negative moment values, joint angular displacement and joint angular impulse were calculated in all three planes for the ankle, knee and hip for the duration of one stride. Peak moments and peak angles for each trial within each condition were averaged to yield the average positive and negative peak values by condition for each participant. Positive and negative angular impulses were calculated by integrating each joint moment curve over the duration of one stride. As changes in gait mechanics can be either in the duration or magnitude of a moment of force, angular impulse will better reflect these two changes than will peak moment values alone (DeVita et al., 1998). Angular impulse values for each trial were averaged to yield the positive and negative angular impulse by condition for each participant. Time integrated EMG data, a measure of the total muscle activity, for each of the muscles tested were averaged within each condition.

The relative positioning of the brace on the leg throughout the gait cycle was determined by comparing the position of the hinge with respect to the knee joint centre. To ensure consistency in the offset between the aligned and misaligned conditions, hinge position waveforms were graphically compared in each condition. Difference scores were calculated by subtracting the misaligned waveform from the aligned waveform for each data point throughout the gait cycle. The mean difference and its standard deviation were calculated from the series of difference scores.

Separate one-way repeated measures ANOVAs, with significance levels of 0.05 were used to determine if peak positive and negative moments in each of the three planes for the ankle, knee and hip were significantly different among the bracing conditions and between the two types of braces. If Mauchly's test, which was run concurrently, indicated a violation of the sphericity assumption, the Huyhn-Feldt correction was used to correct for the violation. If a significant main effect was found, six follow-up one-way repeated measures ANOVAs were used to determine between which comparisons the differences existed. The Bonferroni correction was used, whereby the experiment-wise alpha level of 0.05 was reduced to 0.01 to account for the six additional pair-wise comparisons. Further, one-way repeated measures ANOVAs were used in similar analyses with the positive and negative angular impulses, peak positive and negative joint angular positions and integrated EMG data.

CHAPTER IV

RESULTS

Temporospatial Gait Parameters and Brace Migration Data

Average speed and stride length variables while walking with and without each functional knee brace as well as during misaligned conditions were quite consistent (Table 4.1). With respect to the aligned condition, the amount of migration along the anteroposterior and vertical LCS axes was also quite consistent between braces; the difference in brace position in the mediolateral direction was negligible. For the shell brace, the disparity was 0.6 cm ± 0.15 in the posterior direction and 2.2 cm ± 0.07 down the longitudinal axis of the leg. For the soft shell brace, the difference was 0.6 cm ± 0.23 in the posterior direction and 1.7 cm ± 0.05 down the longitudinal axis of the leg. Neither brace, during the aligned condition, was able to accurately match the position of the knee joint during the gait cycle, with average absolute differences of 0.30 cm ± 0.21 and 0.48 cm ± 0.41 for the shell brace and soft shell brace, respectively (Figure 4.1).

Table 4.1 Mean Velocity and Stride Length Characteristics.

Variable	No-BR	A-SL	M-SL	A-SSL	M-SSL
Velocity (m/s)	1.46	1.43	1.46	1.43	1.43
	(0.18)	(0.18)	(0.18)	(0.17)	(0.19)
Stride length (m)	1.57	1.53	1.53	1.56	1.54
	(0.09)	(0.09)	(0.10)	(0.11)	(0.11)

Note: Standard deviations are reported within parentheses. No-BR is the unbraced condition; A-SL is the aligned shell brace condition; M-SL is the misaligned shell brace condition; A-SSL is the aligned soft shell brace condition; M-SSL is the misaligned soft shell brace condition.

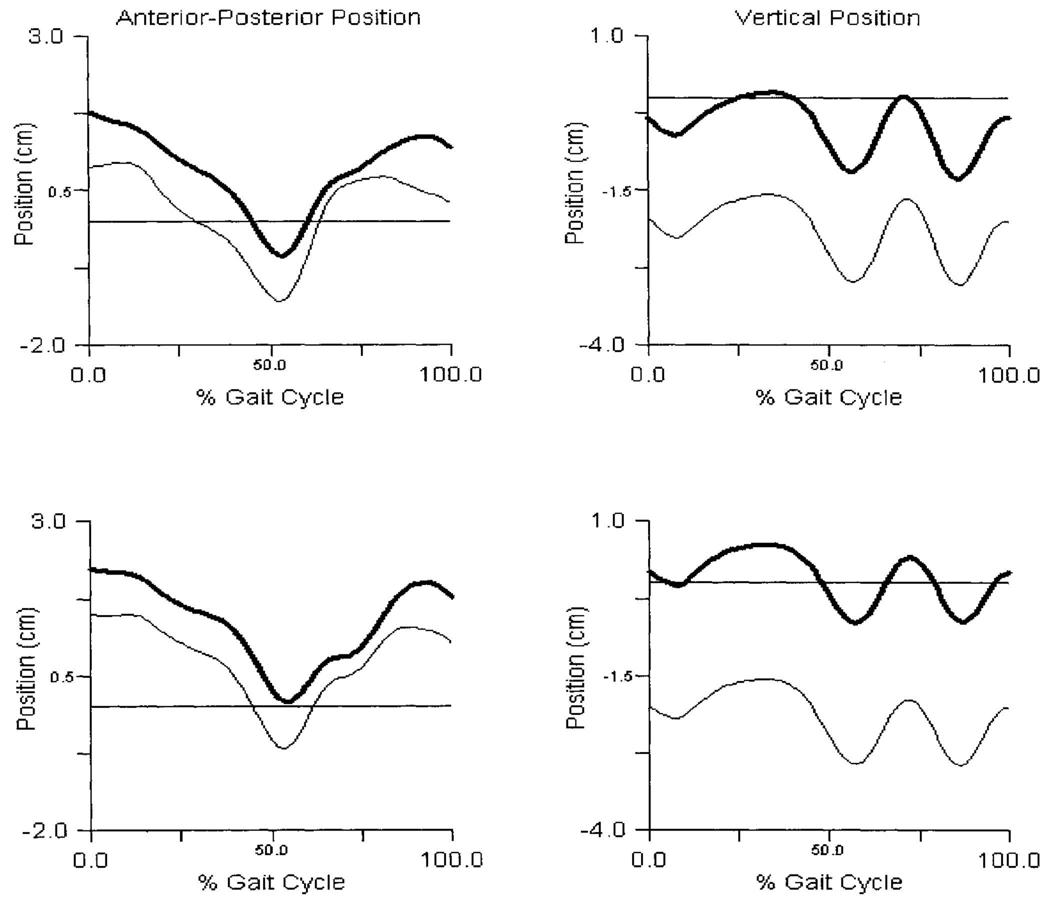


Figure 4.1 Brace position data. Anterior-posterior (left) and vertical (right) position of the brace hinge with respect to the knee joint centre (horizontal line at 0.0) for soft shell brace (top) and shell brace (bottom). Aligned condition is displayed in bold.

Kinetic Data

Joint Angular Impulse

Positive and negative angular impulse values showed consistency across conditions (Table 4.2). The initial one way repeated measures ANOVA failed to show any significant differences at the ankle and knee. There was, however, a significant main effect for the hip abduction moment, though follow-up repeated measures ANOVAs failed to find any differences between meaningful comparisons.

Table 4.2 Mean angular impulse values for the unbraced condition (No-BR); the aligned shell brace condition (A-SL); the misaligned shell brace condition (M-SL); the aligned soft shell brace condition (A-SSL); the misaligned soft shell brace condition (M-SSL).

Impulse (N.m.s/kg)	No-BR	A-SL	M-SL	A-SSL	M-SSL	*P*	η_p^2
Ankle Plantar	0.63 (0.10)	0.65 (0.07)	0.64 (0.08)	0.65 (0.09)	0.66 (0.08)	.450	.104
Knee Extension	0.12 (0.07)	0.12 (0.06)	0.12 (0.04)	0.14 (0.09)	0.13 (0.08)	.643	.073
Knee Flexion	0.13 (0.04)	0.12 (0.04)	0.11 (0.04)	0.11 (0.05)	0.13 (0.06)	.372	.121
Knee Abduct.	0.14 (0.05)	0.13 (0.04)	0.14 (0.04)	0.14 (0.05)	0.15 (0.05)	.160	.181
Knee Int. Rotation	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.01 (0.02)	0.01 (0.02)	.641	.056
Knee Ext. Rotation	0.01 (0.01)	0.01 (0.01)	0.01 (0.01)	0.02 (0.02)	0.02 (0.01)	.219	.160
Hip Flexion	0.10 (0.02)	0.08 (0.03)	0.09 (0.04)	0.09 (0.02)	0.08 (0.02)	.196	.167
Hip Extension	0.26 (0.06)	0.29 (0.09)	0.28 (0.11)	0.27 (0.09)	0.28 (0.06)	.847	.041
Hip Abduction	0.30 (0.57)	0.29 (0.50)	0.27 (0.49)	0.25 (0.49)	0.28 (0.48)	.018*	.302

Note: All measures are reported in N.m.s/kg. Standard deviations are reported within parentheses. Probability values and effect sizes (η_p^2) are reported from the initial ANOVA. * signifies a significant main effect for the initial ANOVA

Peak Joint Moments

Peak joint moment values at the knee and hip also showed consistency across conditions (Table 4.3). Trends in the data included a reduced peak hip abduction moment and peak knee flexion moment in both the aligned shell and aligned soft shell brace condition in comparison to the unbraced condition (Figure 4.3; Figure 4.4). A significant main effect for the ankle plantarflexor moment was found and the Huynh-Feldt correction was used to account for a violation of the sphericity assumption. Follow-up repeated measures ANOVAs showed that participants walked with a reduced peak plantarflexor moment in the aligned shell brace condition in comparison to the unbraced condition $F(1,9)=15.23$, $p<.01$, $\eta_p^2=.63$ (Figure 4.2).

Table 4.3 Mean peak moment values for the unbraced condition (No-BR); the aligned shell brace condition (A-SL); the misaligned shell brace condition (M-SL); the aligned soft shell brace condition (A-SSL); the misaligned soft shell brace condition (M-SSL).

Moment (Nm/kg)	No-BR	A-SL	M-SL	A-SSL	M-SSL	*P*	η_p^2
Ankle Plantar	-2.04 (0.22)	-1.97 (0.21)	-1.95 (0.25)	-2.01 (0.20)	-2.03 (0.23)	.020*†	.329
Knee Extension	0.62 (0.30)	0.58 (0.25)	0.60 (0.19)	0.63 (0.34)	0.56 (0.32)	.838	.043
Knee Flexion	-0.37 (0.15)	-0.30 (0.15)	-0.29 (0.15)	-0.31 (0.18)	-0.34 (0.18)	.271	.145
Knee Abduct.	-0.37 (0.11)	-0.34 (0.11)	-0.38 (0.12)	-0.38 (0.11)	-0.39 (0.10)	.249	.151
Knee Int. Rotation	0.06 (0.04)	0.07 (0.04)	0.07 (0.03)	0.06 (0.04)	0.06 (0.05)	.230	.157
Knee Ext. Rotation	-0.07 (0.03)	-0.06 (0.03)	-0.06 (0.04)	-0.07 (0.04)	-0.08 (0.04)	.104	.208
Hip Flexion	0.50 (0.07)	0.47 (0.10)	0.50 (0.12)	0.48 (0.09)	0.46 (0.10)	.798	.049
Hip Extension	-1.32 (0.38)	-1.33 (0.35)	-1.34 (0.43)	-1.24 (0.23)	-1.24 (0.31)	.272	.145
Hip Abduction	-0.96 (0.19)	-0.87 (0.14)	-0.90 (0.19)	-0.91 (0.16)	-0.91 (0.13)	.125	.197

Note: All measures are reported in Nm/kg. Standard deviations are reported within parentheses. Probability values and effect sizes (η_p^2) are reported from the initial ANOVA.
* signifies a significant main effect for the initial ANOVA.
† signifies a significant difference between the aligned shell brace and the unbraced condition.

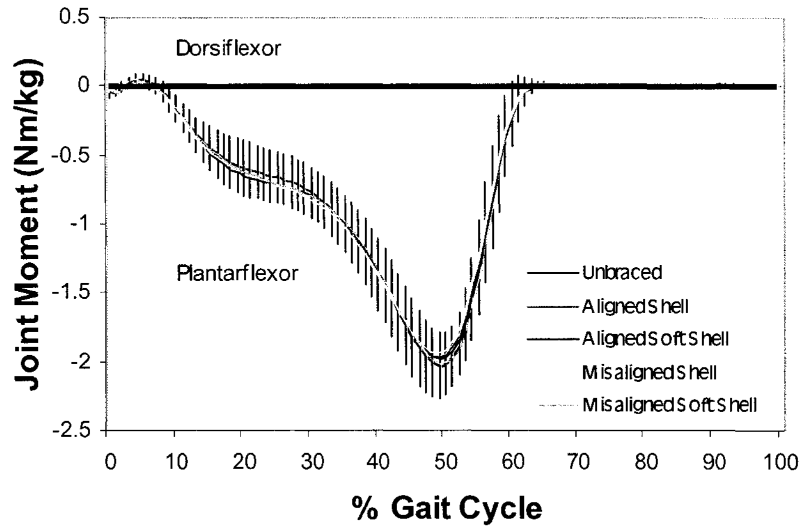


Figure 4.2 Comparison of joint moments (±1 S.D.) for all conditions for the ankle sagittal plane.

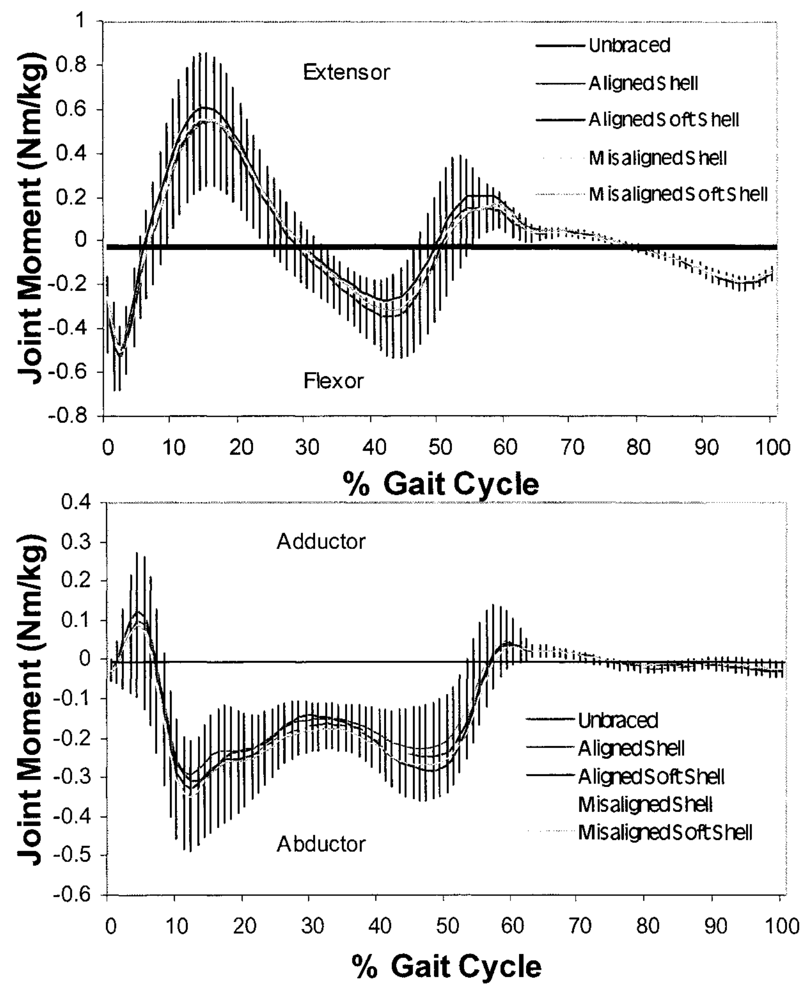


Figure 4.3 Comparison of joint moments (±1 S.D.) for all conditions for the knee sagittal plane (top) frontal plane (bottom).

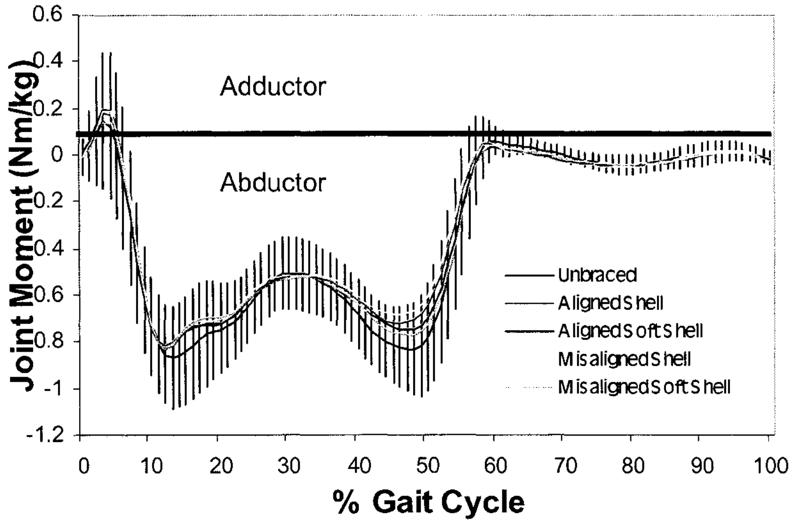


Figure 4.4 Comparison of joint moments (±1 S.D.) for all conditions for the hip frontal plane.

Kinematic Data

Peak Joint Angles

Initial one way repeated measures ANOVAs revealed main effects for all peak joint angles at the knee, with the exception of the peak extension angle. Significant main effects were also found for the peak hip flexion angle (Table 4.4). Additional main effects were found for the peak ankle inversion angle, $F(2.01,18.10)=4.65$, $p<.05$, $\eta_p^2=.34$, the peak ankle eversion angle, $F(4,36)=4.06$, $p<.05$, $\eta_p^2=.31$ and for the peak ankle external rotation angle $F(2.05,18.41)=12.10$, $p<.01$, $\eta_p^2=.57$. The sphericity assumption was violated for the peak ankle inversion and external rotation angles and for the peak knee adduction, abduction and internal rotation angles. The Huynh-Feldt correction was used to adjust for these violations. Trends in the data included a reduced ankle plantarflexion angle in the aligned shell and aligned soft shell brace conditions in comparison to the unbraced condition (Figure 4.5)

Follow-up repeated measures ANOVAs indicated that the knee reached a significantly less flexed state in the aligned shell brace condition, $F(1,9)=21.43$, $p<.01$, $\eta_p^2=.70$ and in the aligned soft shell brace condition, $F(1,9)=23.95$, $p<.01$, $\eta_p^2=.73$ as compared with the unbraced condition. In the frontal plane there was significantly more adduction in the aligned shell condition in comparison to the unbraced condition, $F(1,9)=28.76$, $p<.01$, $\eta_p^2=.76$, and, in the transverse plane, significantly less internal rotation in the aligned shell brace condition, $F(1,9)=23.12$, $p<.01$, $\eta_p^2=.72$ as compared to the unbraced condition (Figure 4.6)

Follow-up tests to the initial main effects found for ankle inversion, eversion and external rotation, knee abduction and external rotation, as well as hip flexion failed to reveal significant differences between any meaningful comparisons.

Table 4.4 Mean peak joint angles for the unbraced condition (No-BR); the aligned shell brace condition (A-SL); the misaligned shell brace condition (M-SL); the aligned soft shell brace condition (A-SSL); the misaligned soft shell brace condition (M-SSL).

Angle (deg)	No-BR	A-SL	M-SL	A-SSL	M-SSL	*P*	η_p^2
Ankle	-15.56	-12.68	-11.90	-13.34	-12.21	.065	.282
Plantar	(6.64)	(6.57)	(6.80)	(7.09)	(6.84)		
Knee	-2.07	-3.61	-3.85	-3.22	-2.37	.174	.175
Extension	(6.63)	(7.15)	(6.59)	(7.86)	(7.34)		
Knee	-66.22	-62.63	-62.21	-63.40	-63.35	.002*†‡	.397
Flexion	(8.40)	(6.65)	(7.77)	(8.20)	(8.29)		
Knee	5.28	8.00	8.56	7.92	8.65	.021*†	.381
Adduction	(4.40)	(4.88)	(7.32)	(5.68)	(6.00)		
Knee	-2.10	-1.21	-0.71	-0.73	-0.32	.041*	.283
Abduction	(3.67)	(4.17)	(4.17)	(3.42)	(3.54)		
Knee Int.	3.37	-0.29	1.08	2.08	2.84	.001*†	.430
Rotation	(2.77)	(4.24)	(4.02)	(4.05)	(3.47)		
Knee Ext.	-7.41	-8.01	-6.96	-6.12	-5.10	.032*	.274
Rotation	(3.52)	(4.03)	(3.07)	(3.20)	(2.93)		
Hip	34.47	33.00	32.58	32.65	32.36	.021*	.295
Flexion	(3.62)	(3.73)	(3.81)	(3.59)	(3.98)		
Hip	-6.61	-4.58	-4.67	-5.81	-5.72	.184	.184
Extension	(3.23)	(3.18)	(2.62)	(2.38)	(2.82)		
Hip	7.16	6.09	5.74	6.49	6.31	.255	.149
Adduction	(2.15)	(2.70)	(2.72)	(1.93)	(2.79)		
Hip	-2.64	-2.21	-3.26	-3.61	-3.96	.227	.157
Abduction	(1.79)	(2.26)	(4.53)	(3.11)	(3.79)		

Note: Standard deviations are reported within parentheses. Probability values and effect sizes (η_p^2) are reported from the initial ANOVA.

* signifies a significant main effect for the initial ANOVA.

† signifies a significant difference between the aligned shell brace and the unbraced condition.

‡ signifies a significant difference between the aligned soft shell brace and the unbraced condition.

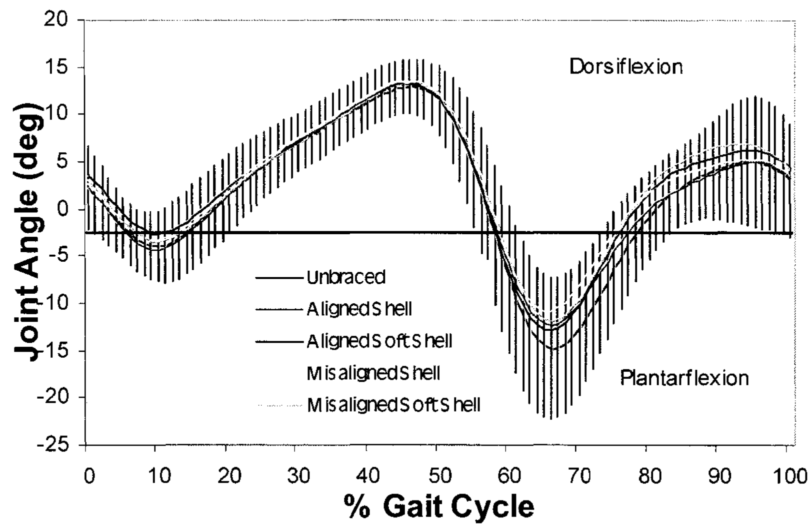


Figure 4.5 Comparison of joint angles (±1 S.D.) for all conditions for the ankle sagittal plane.

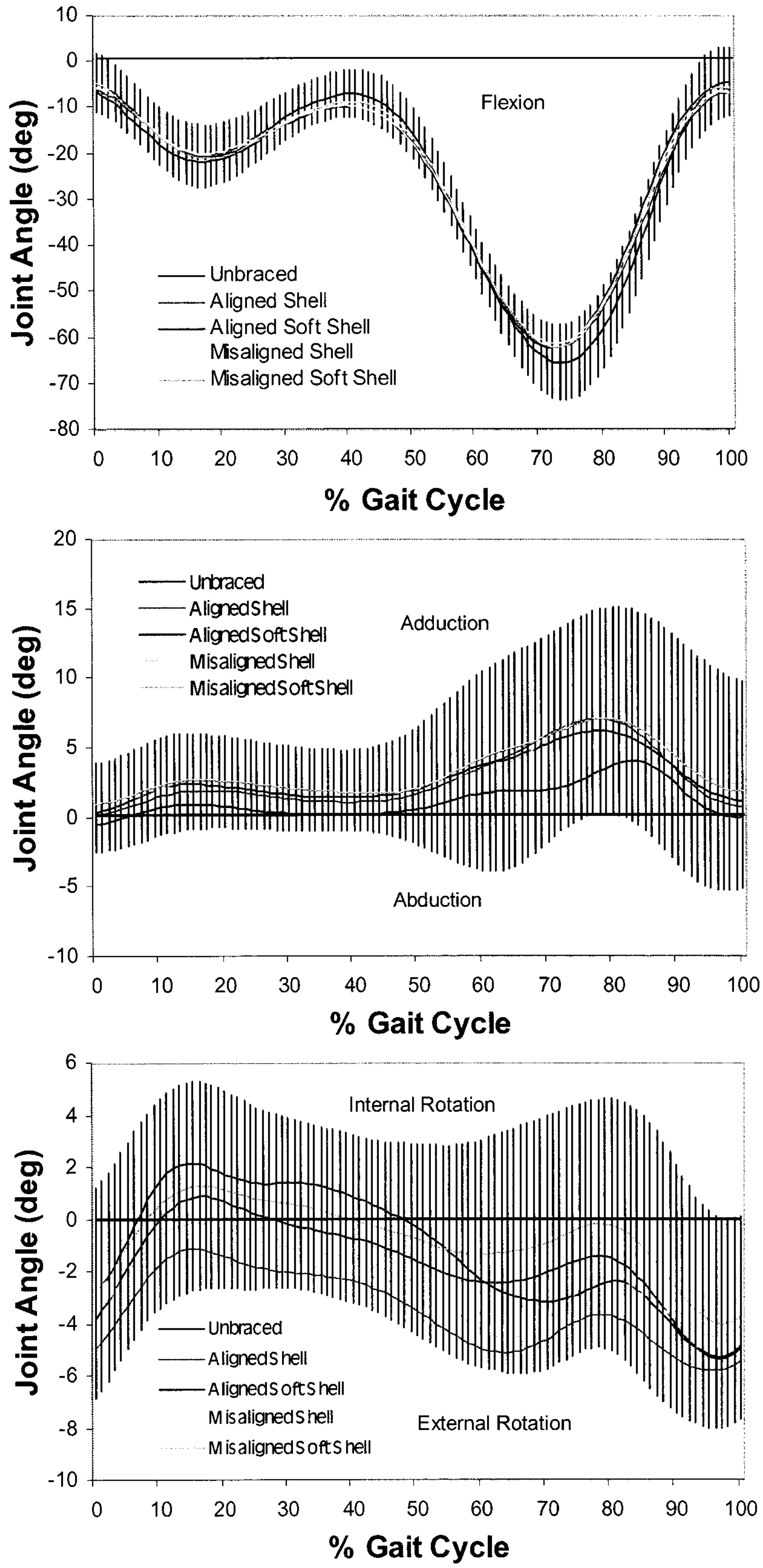


Figure 4.6 Comparison of joint angles (±1 S.D.) for all conditions for the knee. Sagittal plane (top), frontal plane (middle) and transverse plane (bottom).

Electromyographic Data

iEMG

Integrated EMG values of the six aforementioned muscles showed consistency across conditions (Table 4.5). The initial one-way repeated-measures ANOVA failed to show any differences between conditions.

Table 4.5 Mean integrated EMG values for the unbraced condition (No-BR); the aligned shell brace condition (A-SL); the misaligned shell brace condition (M-SL); the aligned soft shell brace condition (A-SSL); the misaligned soft shell brace condition (M-SSL).

Muscle	No-BR	A-SL	M-SL	A-SSL	M-SSL	*P*	η_p^2
Med. Gastroc.	0.12	0.11	0.11	0.12	0.12	.097	.220
	(0.14)	(0.13)	(0.12)	(0.13)	(0.12)		
Lat. Gastroc	0.12	0.11	0.12	0.12	0.11	.299	.138
	(0.07)	(0.07)	(0.07)	(0.08)	(0.07)		
Semitendinosis	0.11	0.09	0.09	0.10	0.09	.240	.164
	(0.07)	(0.07)	(0.05)	(0.09)	(0.06)		
Biceps femoris	0.12	0.10	0.10	0.12	0.09	.287	.144
	(0.10)	(0.09)	(0.10)	(0.12)	(0.08)		
Vast. Medialis	0.10	0.10	0.10	0.12	0.09	.227	.173
	(0.08)	(0.08)	(0.09)	(0.11)	(0.07)		
Vast. Lateralis	0.11	0.10	0.10	0.11	0.09	.164	.202
	(0.08)	(0.05)	(0.06)	(0.09)	(0.06)		

Note: All values are reported in mV.s. Standard deviations are reported within parentheses. Probability values and effect sizes (η_p^2) are reported from the initial ANOVA.

CHAPTER V

DISCUSSION

The current study was designed to determine 1) whether shell or soft shell type functional knee braces alter peak joint moment, angular impulse, peak joint angle or iEMG values during walking in comparison to unbraced walking, 2) if brace migration of either type of brace alters the aforementioned parameters as compared to the corresponding aligned brace position and 3) whether these parameters are altered as a function of brace type in either the aligned or misaligned positions. The following sections will endeavour to examine the brace and migration induced changes during gait at each joint of the lower limb.

Temporospatial Parameters and Brace Migration

Mean velocity and stride length characteristics indicated that neither brace nor alignment altered self-selected walking speed in comparison to the unbraced condition. As Winter (1991) has shown that joint moments can be influenced by cadence, the reported net joint moments in this investigation do not need to be normalized by velocity for correct interpretation. Several authors have investigated the effects of knee braces on performance variables and have reached conflicting conclusions (Houston & Goemans, 1982; Cook, Tibone & Redfern, 1989; Sforzo et al., 1989; Osternig & Robertson, 1993; Greene et al., 2000). While studies of muscle performance during athletic activity have generally indicated a reduction in performance, studies investigating the effects of bracing on measures of speed and agility claim that the brace causes no such reduction. Although the current measures were obtained during walking rather than running, the temporospatial results of this study are in line with those of Greene et al. (2000) who determined that protective knee bracing did not significantly reduce speed or agility during either a 40-yard dash or a four-cone agility drill. Although it is generally accepted that knee bracing does impair muscle performance in the braced limb, there may be underlying mechanical or physiological compensatory strategies that allow individuals to maintain a walking or running speed similar to those measured without the brace. Such strategies could involve alterations in the mechanics of the contralateral unbraced limb, which was not included in the analysis. Moreover, walking may not be a strenuous enough activity for the brace to limit stride parameters.

The amount of imposed distal brace migration in both the shell and soft shell brace was quite consistent between braces and between participants. This allows for a direct comparison of bracing conditions without having to account for the possible confounding effects of differences in migration distances. It was interesting to note that neither brace, in its aligned position, was able to accurately mirror the position of the knee joint centre during the gait cycle. Both the shell brace and soft shell brace hinges followed essentially the same cyclic pattern of vertical (z-axis) migration, where the hinge was located at a specific point at heel-strike, subsequently became offset in either positive or negative direction, and returned to the same position at the successive heel-strike – only at finite instances were the hinges aligned with the knee joint centre (Figure 4.1). It was also interesting to note that, in relation to the aligned position, both braces in their misaligned position followed a similar pattern of offset, though were shifted distally and posteriorly by a predetermined distance. Care should be taken in comparing the position of the brace to the knee joint centre as calculated with videography, as possible errors may persist in the calculation of the joint centre from anatomical marker data due to the sensitivity of the system. Moreover, the effect of joint surface motion was not included in the calculation of the joint centre.

Although this study used videography to measure the hinge position relative to the knee joint centre, the results from the current study are in parallel to those of Blauth, Ulrich and Hahne (1990), who used radiographic imaging to determine that the hinge rarely coincides with the knee axis, with differences of 10 to 50 mm being noted. The migration data from the current study are also in agreement with the results of Lamontagne, Singer and Xu (2003), who assumed that a soft shell brace acted as a rigid body and determined its kinematic pattern during cycling. This study concluded that there was a persistent offset between the kinematic pattern of the knee and of the brace, but this pattern did not worsen over time. Though most other studies indicate that brace migration does worsen over time (Brownstein, 1998; Rast 2000; Wojyts & Huston, 2001), the lack of this finding in the current and aforementioned studies may be due to the fact that the extent of migration was measured during the low-impact activities of walking and cycling.

Kinetics, Kinematics and EMG

The ankle plantarflexor angular impulse values reported in this investigation were not in agreement with the results of DeVita et al. (1996), who showed an increase in plantarflexor angular impulse when using a brace in comparison to normal unbraced walking. The results of the present study indicated that while there were no differences in plantarflexor angular impulse values, there was a reduction of the peak plantarflexor moment at toe-off in the aligned shell condition in comparison to the unbraced condition (Figure 4.2), which may be a contributing factor leading to the reduced peak ankle plantarflexor angle observed following toe-off (Figure 4.5). This finding indicates that while the amplitude of the moment of force was reduced in the aligned shell condition, the total contribution of the moment toward producing the forward propulsive force remained unchanged. This finding is supported by the electromyographic data, which illustrate that the total amount of muscular activity in both the medial and lateral heads of the gastrocnemius also remained stable between conditions.

One possible explanation regarding the reduced peak moment in the aligned shell condition pertains to the effect of the rigidity of the brace and the load distribution over the gastrocnemius. In studies of counterforce bracing on wrist and forearm muscle function, Groppel and Nirschl (1986) and Anderson and Rutt (1992) found that muscle function was compromised because compression over the musculotendinous unit impeded the expansion of the muscles, which inhibited tendon movement and subsequent force production. This explanation also seems plausible in the current study, as in the soft shell brace conditions, where the applied force of the brace is more equally distributed over the shank and the muscle could expand more freely, the peak moment values were not significantly different from the unbraced condition.

It is known that the gastrocnemius contributes to both the ankle plantarflexor moment and the knee flexor moment during mid- to terminal stance (Winter, 2005). While the gastrocnemius is capable of acting as a knee flexor at a multitude of knee and ankle joint angles, Li, Landin, Grodesky and Myers (2002) have found that the magnitude of the knee flexor moment caused by the gastrocnemius is dependent on both the position of the knee and ankle joints. The non-significant trend of a reduced peak knee flexor moment in the aligned shell brace condition in comparison to the unbraced condition may also be closely tied to the gastrocnemius, particularly since the sagittal plane angular position of the ankle and knee joints were comparable across conditions during mid- to terminal stance (Figure 4.5; Figure 4.6).

A number of authors have indicated that the hamstrings, the contribution of the hamstrings to the knee flexor moment and the overall magnitude of the knee flexor moment may be helpful in reducing anterior tibial translation and ACL strain (Branch et al. 1989; DeVita et al. 1992; DeVita et al. 1996), as the muscle contribution to the flexor moment exerts a posteriorly directed force on the tibia. Irrespective of the relative contributions of the hamstrings or gastrocnemius to the net joint moment, a reduction in the peak knee flexor moment observed in both the shell and soft shell brace conditions may be deemed to be negative, as it may limit the protective effect on the ACL. With this being stated, however, it seems unlikely that a reduction in the peak flexor moment at the knee would have much bearing on the amount of anterior tibial translation or ACL strain within the scope of this study, particularly since there were no significant differences in gastrocnemius or hamstring muscle activity or in ankle plantarflexor or knee flexor angular impulses across conditions. Furthermore, Shelburne et al. (2004a) and Shelburne et al. (2004b) have found that the peak ACL force occurs at 15% of the gait cycle (contralateral toe-off) and corresponds to the increase in quadriceps force during this period. In the current study, there were no changes in quadriceps iEMG values or in the extensor angular impulse or peak moment, which may apply an anterior shear force to the tibia. Moreover, the reduction of the peak flexor moment occurs from approximately 30% to 50% of the gait cycle – a time where the application of a posteriorly directed force on the tibia is not as necessary, as predicted ACL strain values drop to half of the peak value due to a reduction in anterior shear loads (Shelburne et al. 2004b; Shelburne et al. 2004a).

Kinematic analysis of the knee joint revealed that, in comparison to the unbraced condition, there was a reduced peak knee flexion angle in both the shell and soft shell brace conditions (Figure 4.6). As knee and hip sagittal plane moment data are nearly identical across conditions at peak knee flexion (Figure 4.3: Figure 4.4), the reduced peak flexion angle is most likely due to the stiffness that the brace imparted to the knee, reducing its ability to flex as it would without the brace. Peak flexion at the knee occurs during swing, a time where the ACL is only slightly loaded due to reduced shear force caused by the quadriceps (Shelburne et al., 2004b). From this, it is believed that there is little chance the amount of flexion during swing has any bearing on the overall safety of functional knee bracing.

The increase in the knee adduction angle in the braced conditions (Figure 4.6), evident throughout the entire gait cycle, may place greater loads on the medial condyle and fibular collateral ligament during stance. The external knee adduction moment caused by the ground reaction force, defined as the moment that tends to adduct the knee during gait, has been shown to increase loads in the medial compartment (Baliunas et al., 2002) and a reduction in external adduction moment has been shown to reduce joint loading without further strain on the ACL (Hewett, Blum, & Noyes, 1997). The internal knee abduction moment – the reaction to the external knee adduction moment – has been found to be a significant predictor of ACL injury, as valgus moments can increase ACL force (Hewett, Myer, Ford, Heidt, Colosimo, McLean, van den Bogert, Paterno, & Succop, 2005). The abductor moments measured in this study, however, were not statistically different between bracing conditions and therefore most likely were not contributing to increased joint loading or injury risk. With this similarity of knee frontal plane moment profiles across conditions (Figure 4.3), it is likely that the brace simply forced the leg into a more adducted position. From the data in this study, it is difficult to determine the role of the brace in sharing the load that would be transmitted to these inert structures. Off-loader braces have been shown to reduce loading of the medial compartment of the knee (Komistek, Mahfouz, Dennis, &

Nadaud, 2005), though it is unclear as to whether the functional braces used in this study would perform the same action. Using approaches similar to Pollo, Otis, Backus, Warren and Wickiewicz (2002) and Singer and Robertson (2004), whereby the brace supports were instrumented with strain gauges, it may be possible to parse out the contribution of the brace in frontal plane joint loading.

Nevertheless, as the knee was more adducted during stance, this may serve to reduce the observed hip abduction moment. Winter (2005) has stated that during stance, the hip abductor pattern is used to prevent the drop of the pelvis against the forces of gravity, which are acting approximately 10 cm medial to the stance hip. A more adducted leg position may act to move the line of action of the gravitational force closer to the hip joint centre, reducing the need for an abductor moment to prevent pelvic drop.

The knee was also in a more externally rotated position throughout the gait cycle, except for the time period immediately before heel-strike, where the knee in both braced conditions approximated the transverse plane angular position of the unbraced knee (Figure 4.6). McLean, Neal, Myers and Walters (1999) have stated that internal axial rotation of the tibia is known to increase loading of the ACL. As the brace reduced internal tibial rotation to a more externally rotated position without corresponding changes in transverse plane knee moments (Table 4.2: Table 4.3), this alone may be considered beneficial in reducing strain on the ACL. Conversely, excessive external tibial rotation has also been implicated in the mechanism of non-contact ACL injury. Due to the relatively small difference in magnitude, however, it is unlikely that the transverse plane knee angles seen in this study would be placing the knee joint in a compromised position.

Overall, the initial hypotheses based on studies by DeVita et al. (1996) and DeVita et al. (1998) do not corresponded with the present data. Whereas these studies indicated that bracing would evoke an increase in sagittal plane extensor angular impulse at the ankle and hip and a reduction in extensor angular impulse at the knee, the current study found no such differences in angular impulse values. These results are substantiated by the stability of iEMG values across conditions. Both braces, however, caused decreases in the peak flexion angle when in their aligned position. As the peak flexion moment occurs during swing, when the ACL is only slightly loaded due to reduced quadriceps shear force, this alteration in gait pattern is not considered a cause for concern. Both braces also failed to alter most of the kinetic parameters indicating that, during walking, there is little risk that bracing will lead to mechanical gait alterations that may expose the ACL to undue strain, which over time may jeopardize the integrity of the ligament.

Regalbuto et al. (1989) have shown proper placement of the hinge to be more important than either brace or hinge type, and speculated that improper placement of the hinge may lead to altered internal joint mechanics or abnormal ligament length patterns. These authors measured force values within the hinge and made inferences regarding its effect on joint mechanics. The current study investigated the effect of brace type and hinge alignment on joint mechanics during walking and found no differences when examining the effect of either brace type or hinge alignment. It is interesting to note that Regalbuto et al. (1989) found that the divergence in force values within the hinge became most pronounced when flexion increased beyond 60 degrees. In the present study, the knee flexion angle reached a maximum value of 66 degrees in the unbraced condition and was reduced in all braced conditions (Table 4.4; Figure 4.6). This factor may be one reason which explains the lack of significant changes in knee moment profiles between conditions. While it was mentioned previously that the rigidity imparted to the knee by the brace may reduce the knee flexion angle, based on the results of Regalbuto et al. (1989), the reduced flexion may prevent abnormal forces in the hinge, which may avoid the development of abnormal moments at the knee. Activities that involve greater knee flexion, especially during weightbearing, may evoke changes in joint moment patterns when the brace is misaligned.

CHAPTER VI

CONCLUSIONS AND RECOMMENDATIONS

This study measured the changes in lower limb mechanics that are caused by functional knee bracing. More specifically, the goal was to determine the effects of brace type and alignment on the three-dimensional angular impulse, peak joint moment, peak joint angular position and iEMG values during walking. Similar to studies by DeVita et al. (1996) and DeVita et al. (1998), rather than assessing the direct mechanical effect of brace type and alignment, it was hoped that this study would shed light on the changes in motor patterns as signified by kinetic, kinematic and electromyographic analysis. The results suggested that while the shell and soft shell type functional knee braces used in this study caused changes in several kinetic and kinematic parameters, these changes most likely do not alter motor patterns sufficiently enough to impose any additional risk to the joint or ACL than would occur during walking without a functional knee brace. Furthermore, misaligning either type of brace does not evoke any changes in joint mechanics that would lead to the conclusion that there is undue risk to any joint of the lower extremity.

Future studies should include running or other activities where the joint range of motion is increased, as this will serve to increase the stresses placed on the lower extremity. Additionally, a longer duration study where measures are taken over multiple strides may be better able to determine the effects of functional knee brace migration on the lower limb. A study addressing the effects of bracing on the contralateral limb should be carried out to determine if there are strategies used to compensate for the restrictions placed on the braced limb. Finally, a study of these effects in ACL-deficient participants should be performed, as an injured sample may be more susceptible to the effects of the brace than a healthy sample.

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APPENDIX A

Joint Moment Data

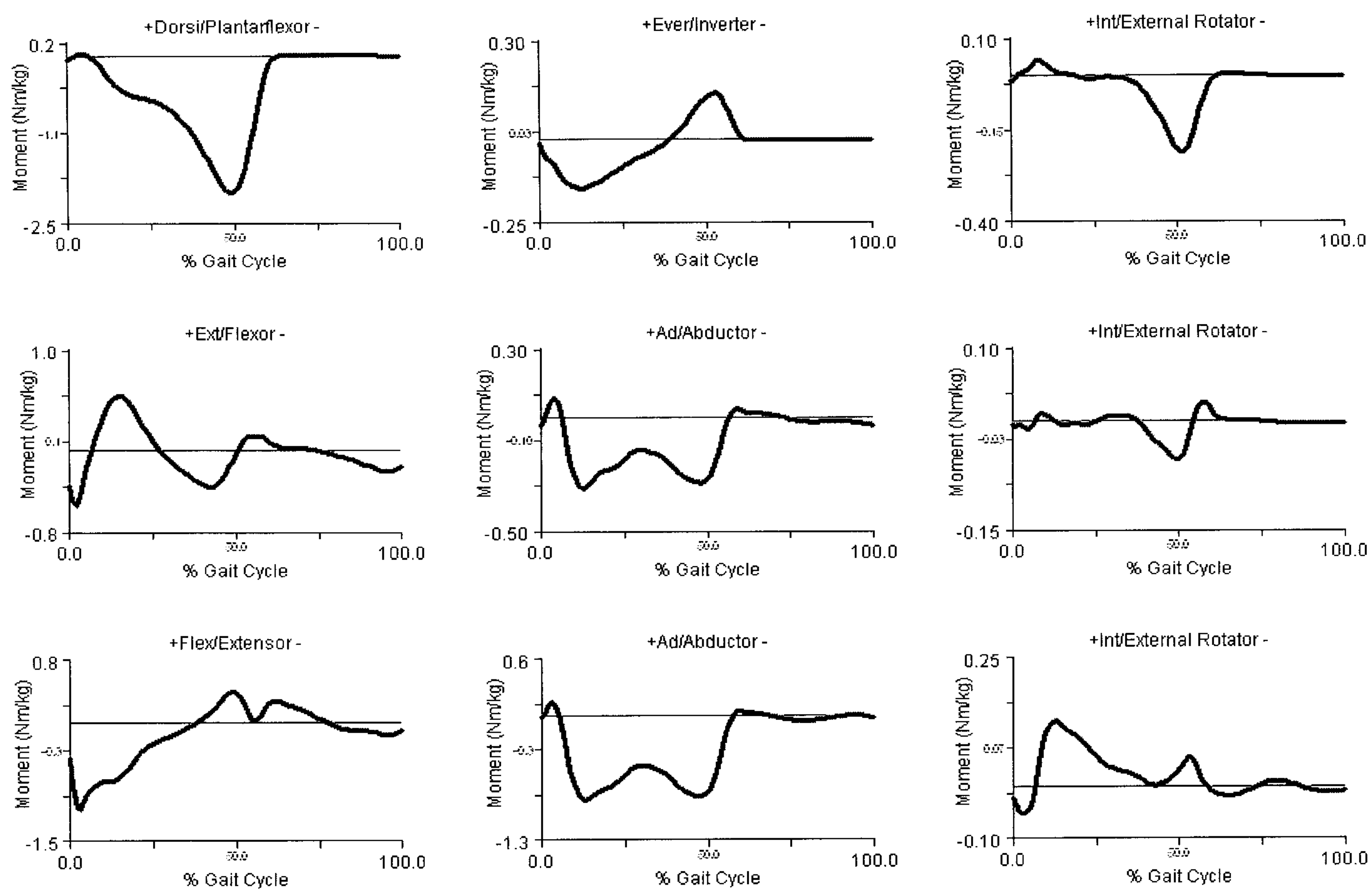

Figure A.1 Mean (+/-1 SD) moments of the ankle (top) knee (middle) and hip (bottom) for the sagittal (left), frontal (middle) and transverse (right) planes for all subjects in the unbraced condition.

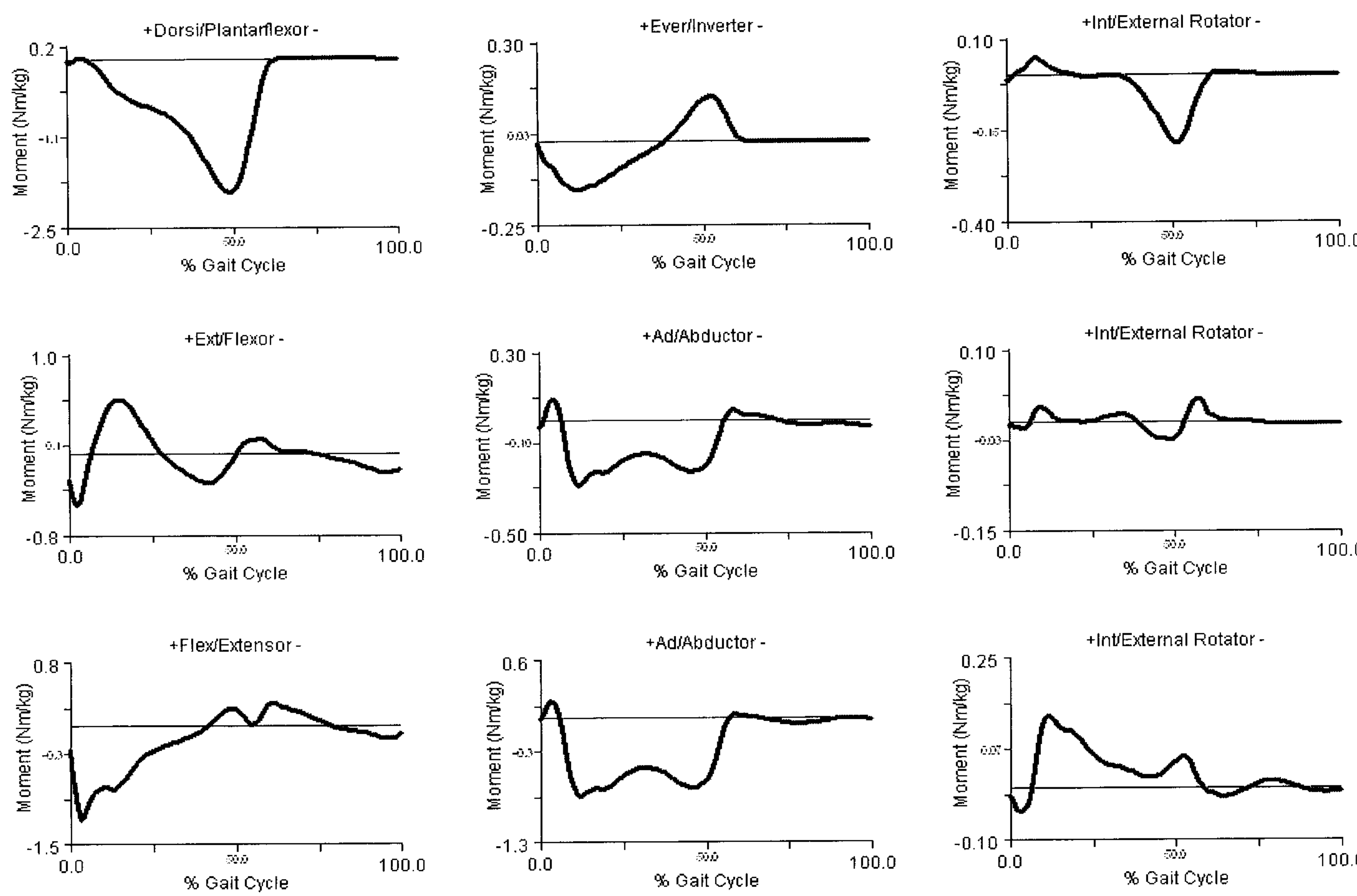

Figure A.2 Mean (+/-1 SD) moments of the ankle (top) knee (middle) and hip (bottom) for the sagittal (left), frontal (middle) and transverse (right) planes for all subjects in the aligned shell brace condition.

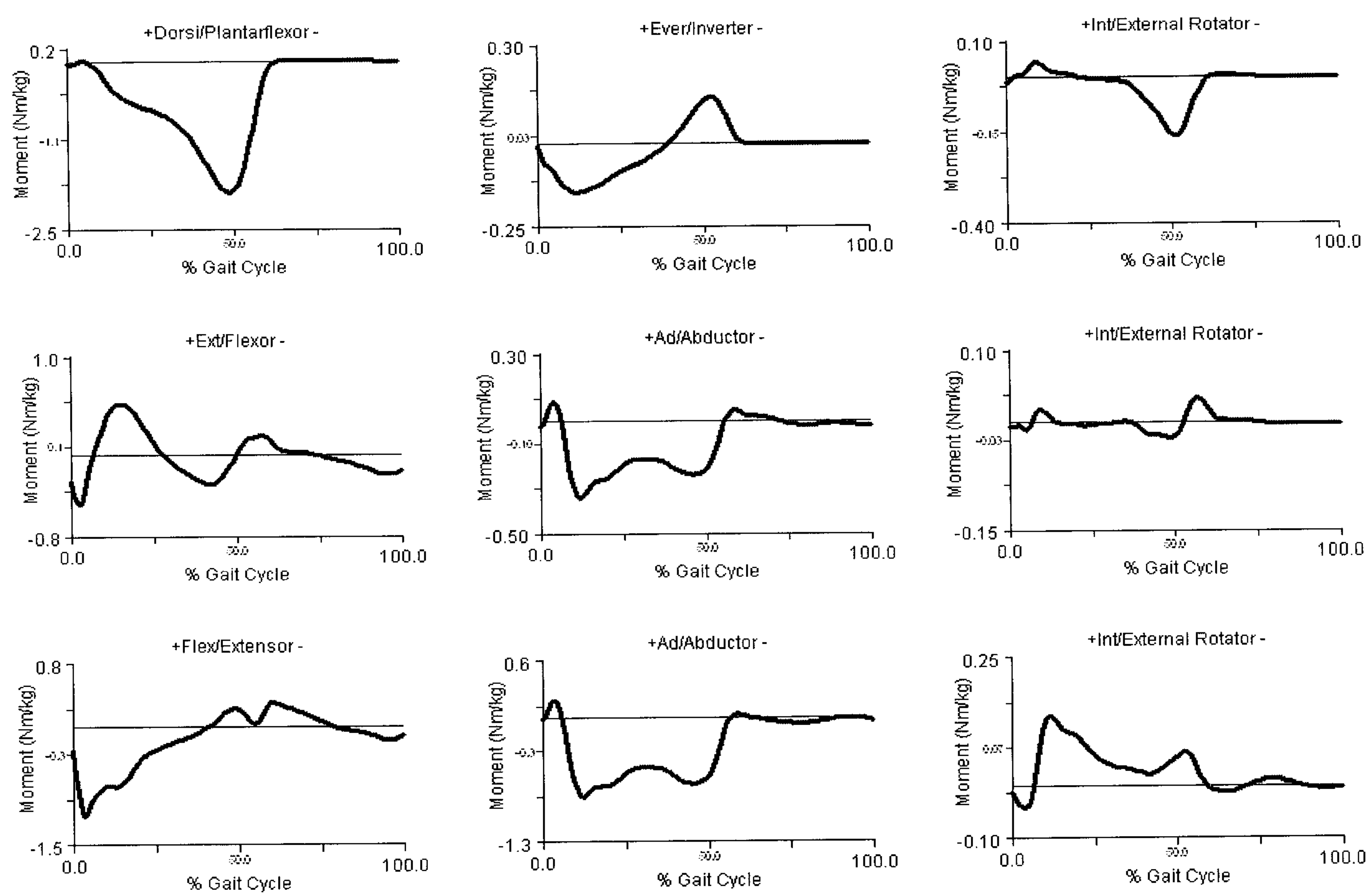

Figure A.3 Mean (+/-1 SD) moments of the ankle (top) knee (middle) and hip (bottom) for the sagittal (left), frontal (middle) and transverse (right) planes for all subjects in the misaligned shell brace condition.

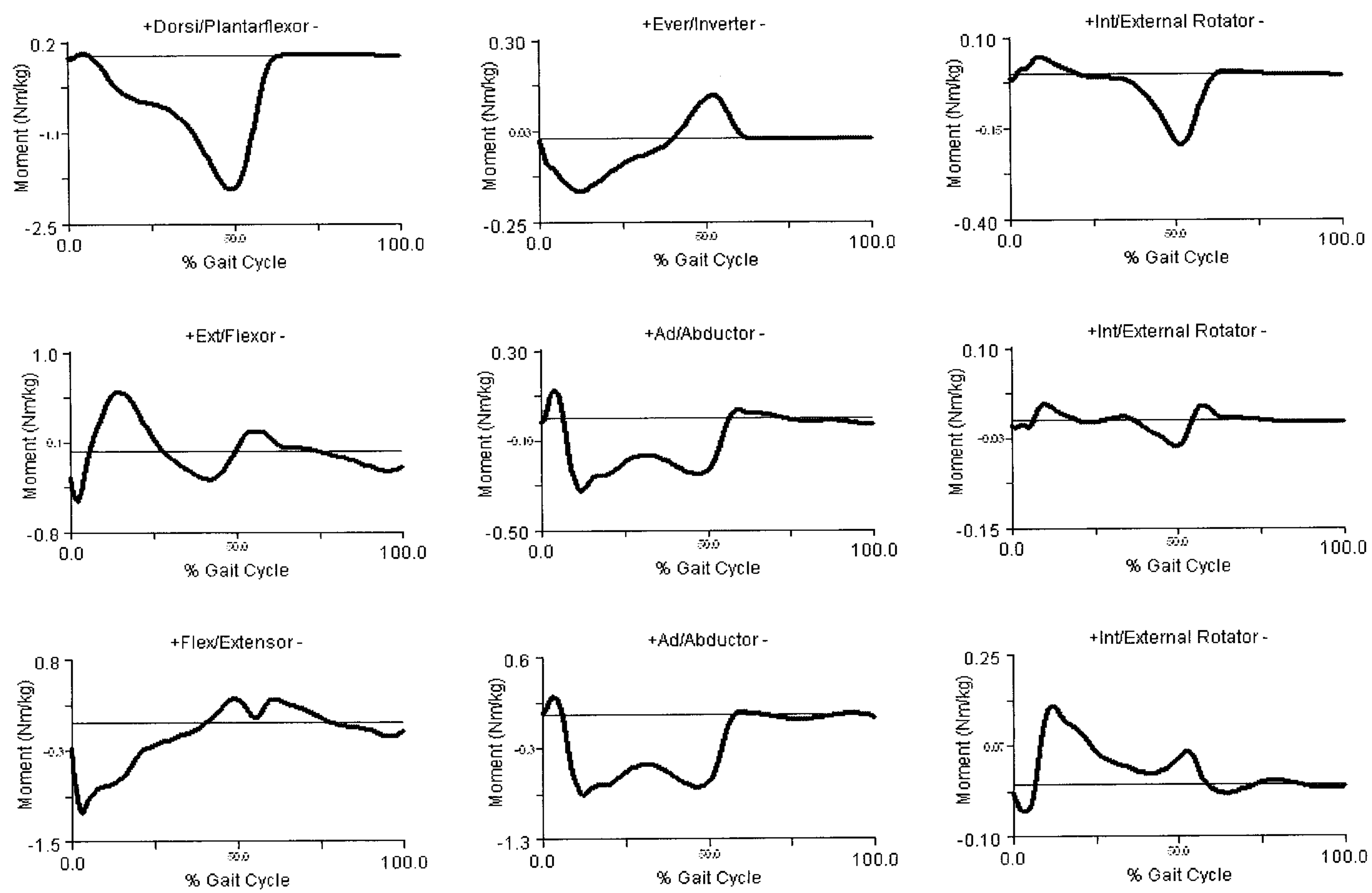

Figure A.4 Mean (+/-1 SD) moments of the ankle (top) knee (middle) and hip (bottom) for the sagittal (left), frontal (middle) and transverse (right) planes for all subjects in the aligned soft shell brace condition.

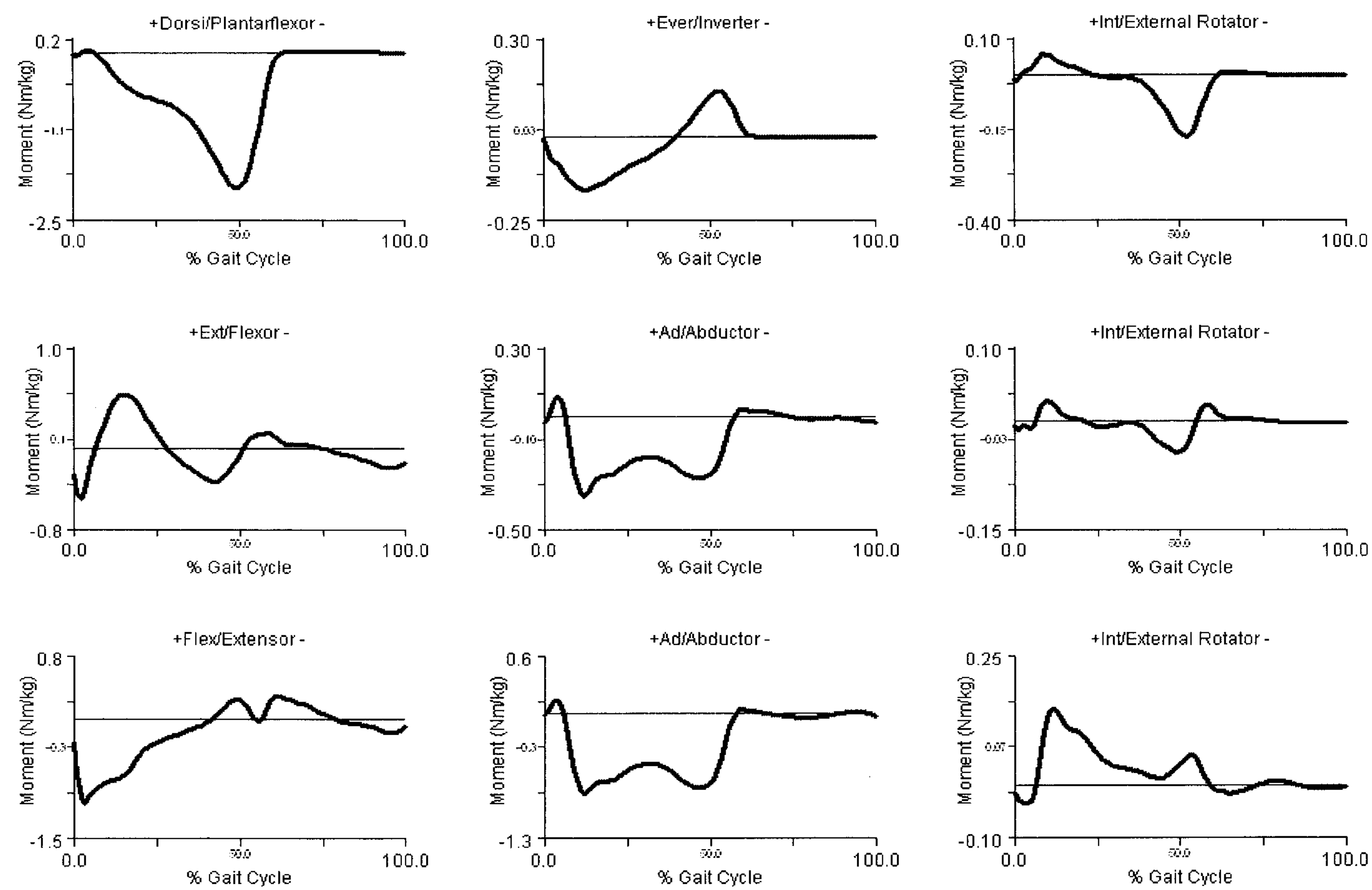

Figure A.5 Mean (+/-1 SD) moments of the ankle (top) knee (middle) and hip (bottom) for the sagittal (left), frontal (middle) and transverse (right) planes for all subjects in the misaligned soft shell brace condition.

APPENDIX B

Angular Displacement Data

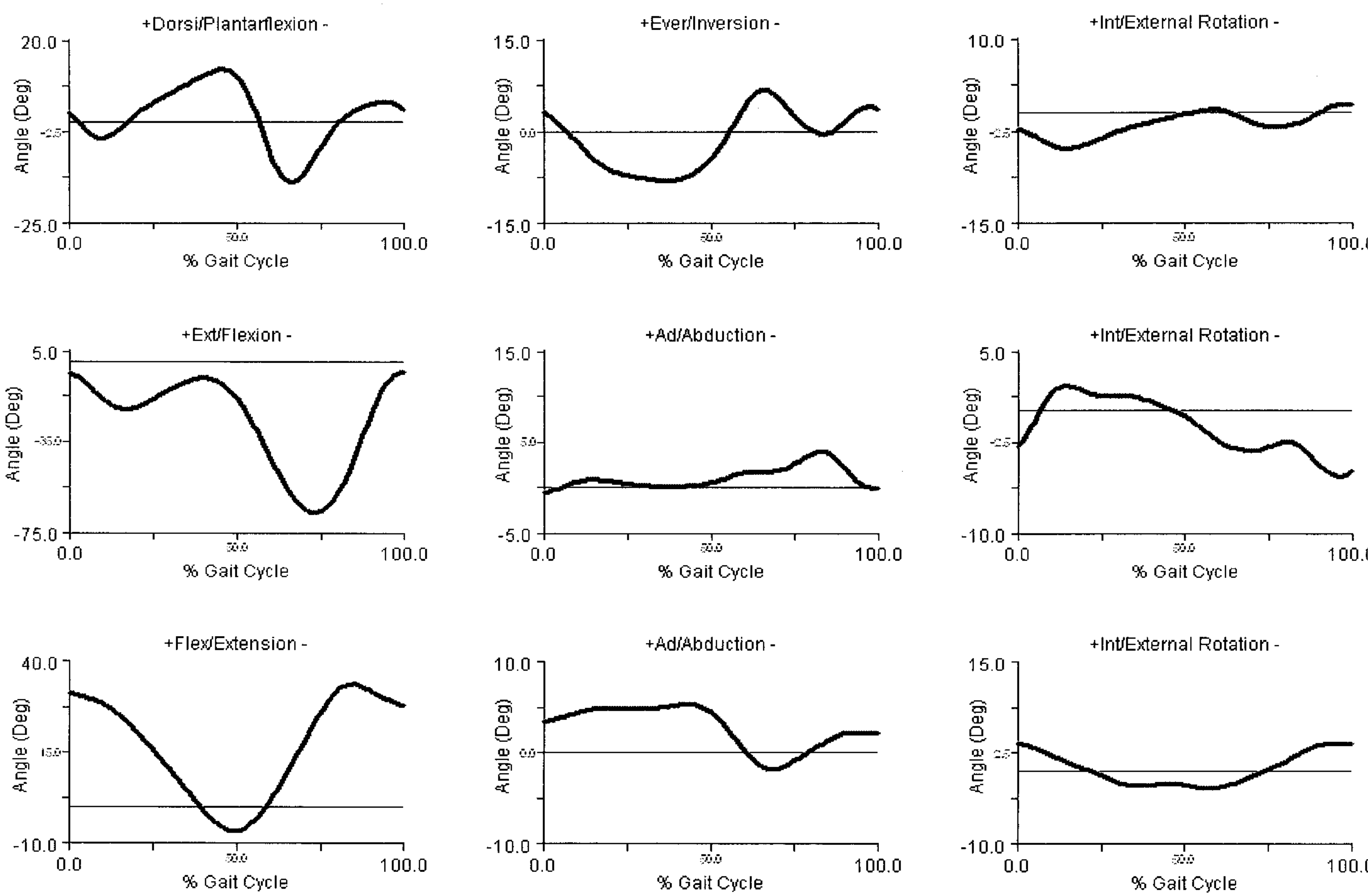

Figure B.1 Mean (+/-1 SD) joint angles of the ankle (top) knee (middle) and hip (bottom) for the sagittal (left), frontal (middle) and transverse (right) planes for all subjects in the unbraced condition.

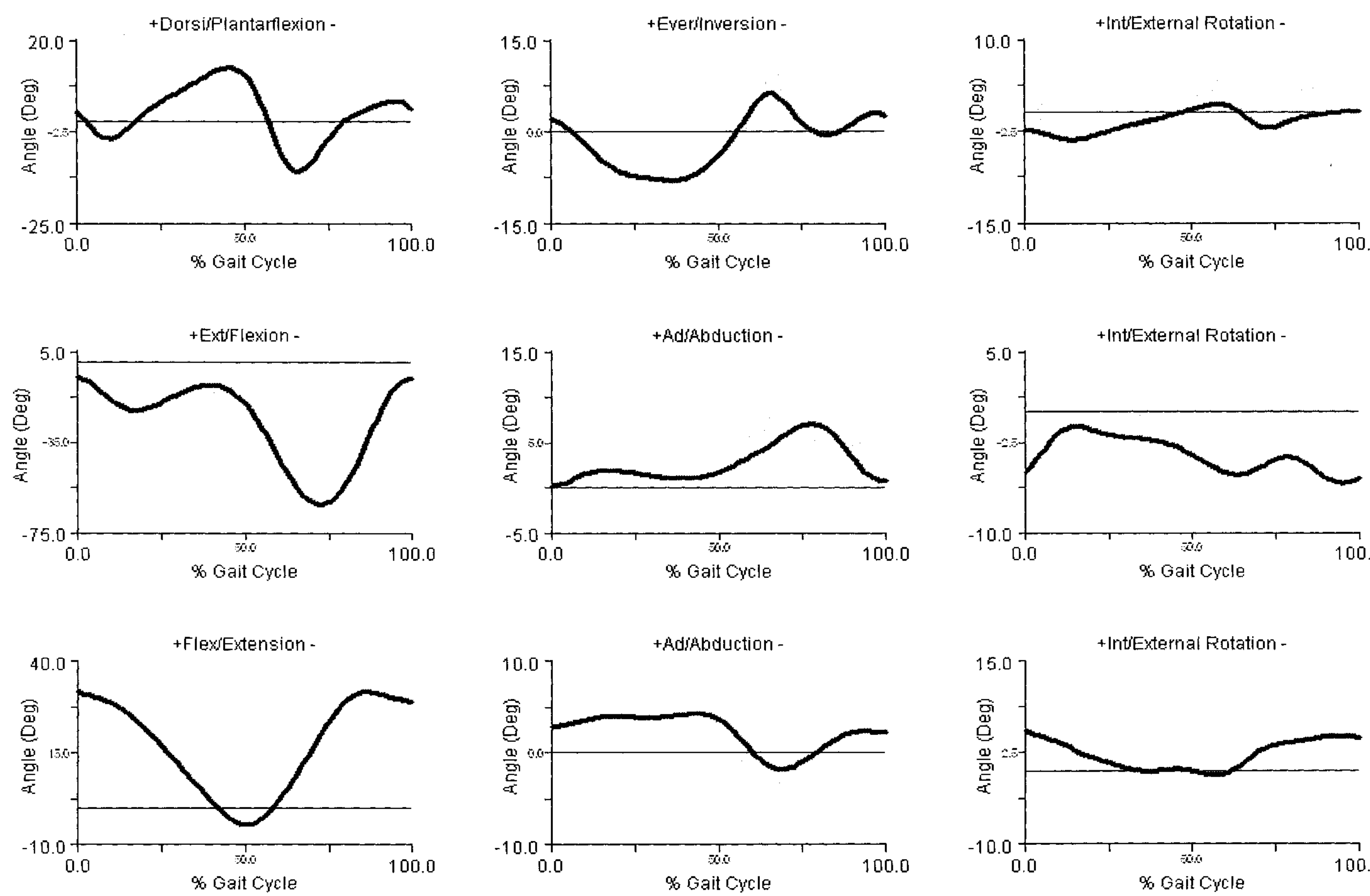

Figure B.2 Mean (+/-1 SD) joint angles of the ankle (top) knee (middle) and hip (bottom) for the sagittal (left), frontal (middle) and transverse (right) planes for all subjects in the aligned shell brace condition.

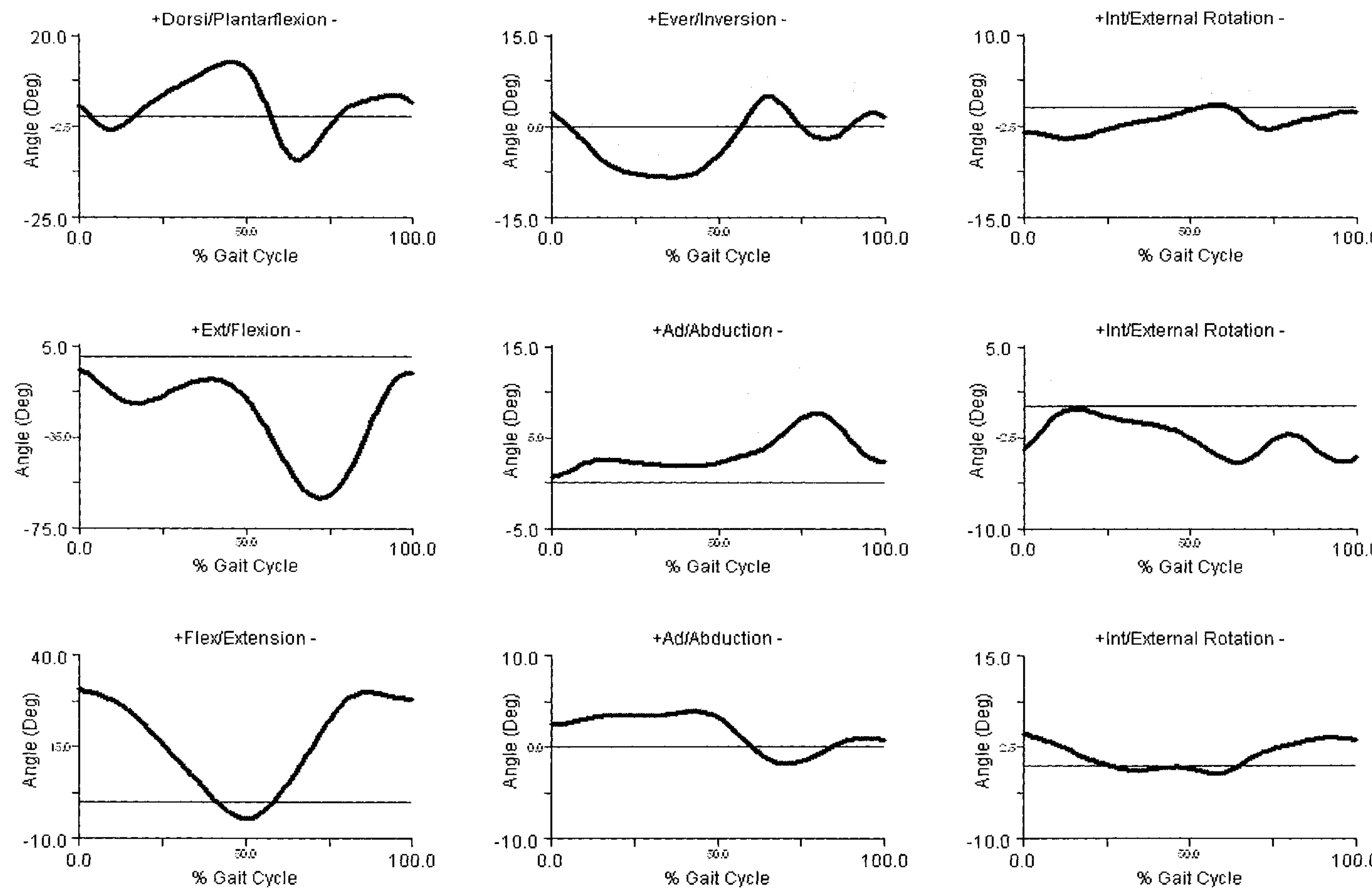

Figure B.3 Mean (+/-1 SD) knee joint angles of the ankle (top) knee (middle) and hip (bottom) for the sagittal (left), frontal (middle) and transverse (right) planes for all subjects in the misaligned shell brace condition.

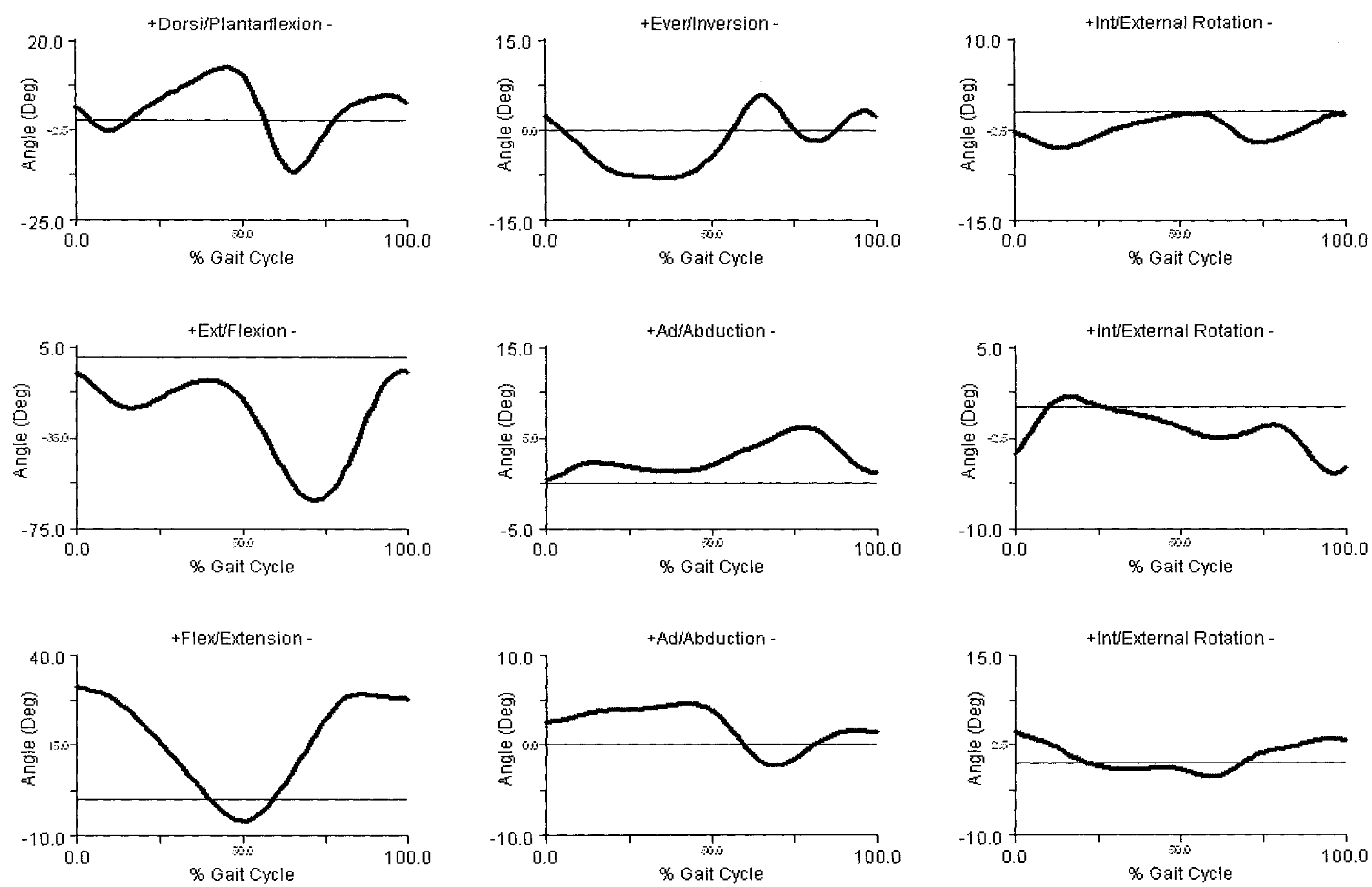

Figure B.4 Mean (+/-1 SD) joint angles of the ankle (top) knee (middle) and hip (bottom) for the sagittal (left), frontal (middle) and transverse (right) planes for all subjects in the aligned soft shell brace condition.

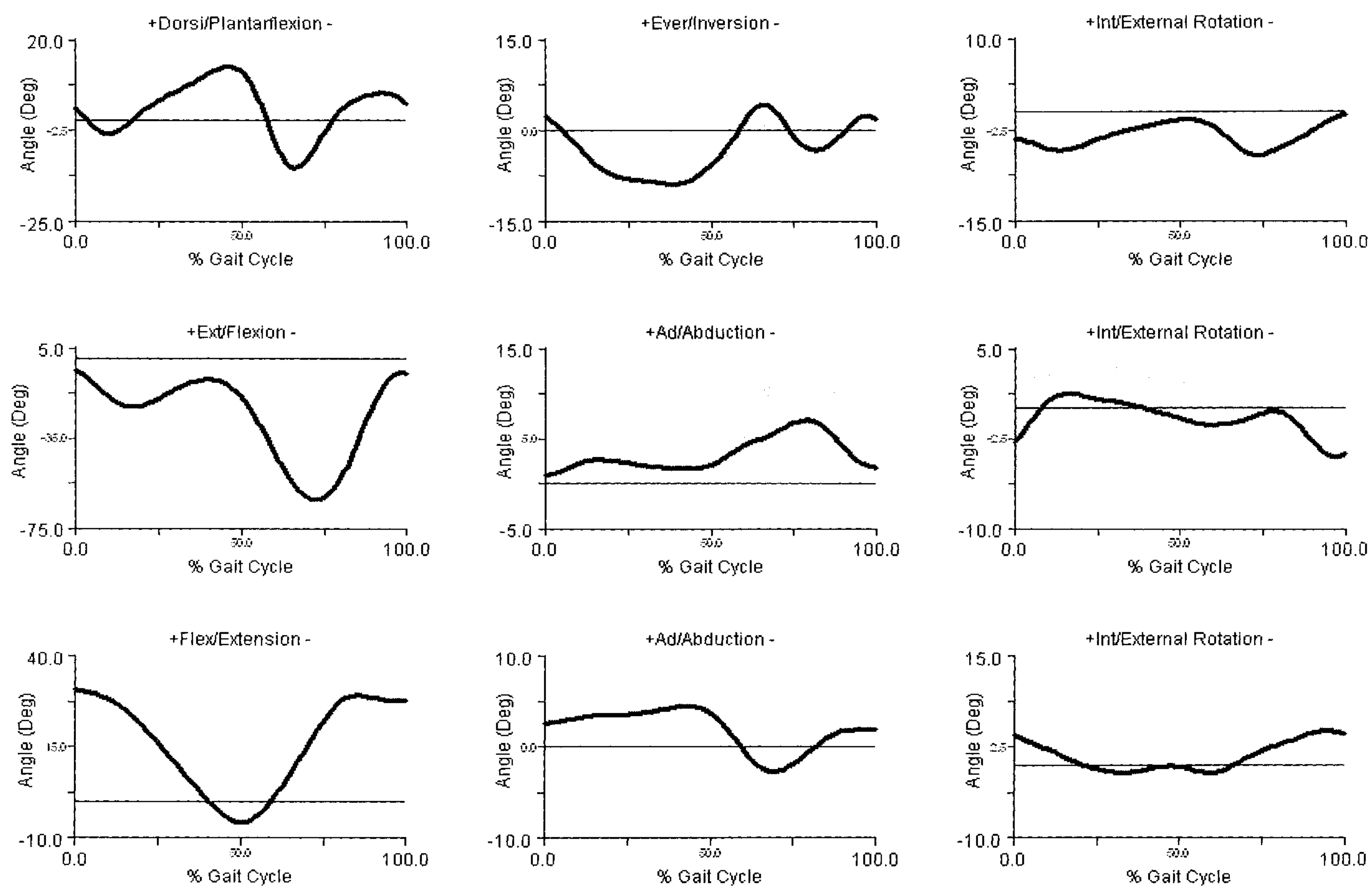

Figure B.5 Mean (+/-1 SD) joint angles of the ankle (top) knee (middle) and hip (bottom) for the sagittal (left), frontal (middle) and transverse (right) planes for all subjects in the misaligned soft shell brace condition.