Saint Xavier University

3D Evaluation of the human gait cycle with respect to prosthetics

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Abstract

Over a million Americans have had limb amputations, and unintentional falls are the number one cause of nonfatal injuries. Amputees are prone to falling injuries due to their unsatisfactory stability while walking on prosthetic limbs. To enhance the quality of life for amputees this research seeks to improve walking ergonomic stability by suggesting possible adjustments to prosthetic design and therapy. Despite increasing affordability and availability of motion capture technology for gait analysis, few studies have utilized this technique to evaluate trans-femoral amputee gait mechanics. Using the motion capture freeware MyCap Studio, a normal, healthy gait cycle is compared to a gait cycle that approximates a distal right leg amputation with a prosthetic. A pseudo leg for able bodied participants is created to represent an amputee’s compromised walk. Qualitative and quantitative data from the motion capture analysis reveal knee and hip landmarks in prosthetic gaits have increased lateral movement compared to natural gait cycles. The study concludes that the amputation and prosthetic limb gait cycle deviates during the push off and pre swing phases primarily due to an absence of an ankle joint, which contributes to overuse muscle fatigue and injury. This contribution to the field of prosthetics is anticipated to aid in future prosthetic development and physical therapy.
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INTRODUCTION

There are currently 2 million American amputees (Ziegler-Graham, 2008). Of those with lower limb amputations, unintentional falls are the number one cause of nonfatal injuries. To enhance the quality of life for amputees this research seeks to improve walking ergonomic stability by suggesting possible adjustments to prosthetic design and physical therapy.

Anatomy of Stability

To better understand the anatomy and mechanical design involved in human hindlimb stability, I investigated whether each muscle in a human’s hindlimb has its own unique contribution towards stability. In other words, do muscles work independently to establish balance?

To investigate this question, extensive cadaver dissection and study was used to determine body movements when “contracting”, or rather pulling, on specific muscles. The two cadavers used were both in their 70’s, one healthy female and one male with a double knee replacement. Due to rigor mortis, certain ranges of motion were restricted, such as dorsiflexion of the ankle, but the use of lab dissector manuals aided in a complete understanding of muscle functions. Research published in scientific journals was also used in conjunction with the cadavers.

Loram (2004) used an ultrasound scanner and automated image analysis to record muscular contractions and changes in muscle length during standing and voluntary swaying. In this paper, subjects were told to stand with legs even to each other at comfortable distances apart and were asked to sway forwards and backwards with knees and hips locked. Passive tissue alone can’t maintain balance but it does help in counteracting gravitational forces on the body. In particular, Loram (2004) found among the standing paradoxical movement data that when
leaning forward, maintaining balance required an increase in ankle torque. An increase in tension activates the muscle and shortens it while the tendon is passively stretched. A laser range finder measuring the ankle’s angle and a foot plate monitoring ankle torque helped in determining that the gastrocnemius and soleus of the lower leg muscles shorten when swaying forward, working together to maintain stability.

Not surprisingly, cadaver dissections and published data show that muscles do not contract independently to maintain stability. In fact, muscles work in groups consisting of flexors and extensors of the hips, knees, and ankles (Abrahams, 2008). Specifically, hip flexion is achieved through the activation of the adductor brevis and adductor longus while the gluteus maximus has control of hip extension. The flexor muscles of the knee consist of the biceps femoris, semimembranosus, semitendinosus, sartorius, and popliteus. The rectus femoris, vastus intermedius, vastus lateralis, and vastus medialis, on the other hand, contribute to extension of the knee. In dorsiflexion at the ankle, made possible by the tibialis anterior, the foot is flexed towards the shin. Plantar flexion, pointing of the toe downwards, is done with the gastrocnemius, fibularis brevis, fibularis longus, plantaris, soleus, and tibialis posterior (Abrahams, 2008). Understanding these muscle groups is imperative when piecing together the functionality represented throughout the gait cycle.

**The Gait Cycle**

Gait refers to the alternating movement of an appendage when moving the body forward, and a gait cycle begins and ends when the same walking event occurs (e.g. the same foot striking the ground)(McGowan, 1999). The walking cycle can be divided into the following two phases: the stance phase begins when a forward limb’s heel touches the ground and ends when the same foot’s toe leaves the ground; the swing phase begins when the same limb is off the ground and
free to move forward to return to the beginning of the stance phase. Each phase is then broken down further into individual stages: initial contact, loading response, mid-stance, push-off, and pre-swing.

During the stance phase of the gait cycle, ankle dorsiflexors and plantar flexors as well as hip and knee flexors and extensors are necessary to compensate for gravitational and produced forces on the body. Initial contact of the heel impacting the ground requires ankle dorsiflexors activation to control the motion of the plantar flexors and balance the external ankle movement. Without this activation, the foot would slap back down to the ground. At this point, the ground reaction force is also in front of the hip, thus causing external movement that flexes the hip. The loading response requires multiple knee extensors to generate tension while also lengthening and absorbing impact. During this phase, the gluteal muscles help the trunk of the body stay erect. Mid-stance, the following stage, involves passive rotation of the center of mass over the ankle. Late in the mid-stance, when the center of mass begins to shift forward, the plantar flexors activate an eccentric contraction. This eccentric action means that the force of the plantar flexors is less than the external force, in this case gravity, making it a stable yet still passive rotation. In push-off, plantar flexors produce tension through shortening and extending the leg to gain the energy lost due to impact of the other leg. In addition, hip flexors activate to counteract the ground reactive force. Between push-off and pre-swing the ankle generates most of its power. This is due to the achilles and calf muscles plantar flexing in order to push off from the ground and enter the swing phase of the gait cycle. Lastly, the pre-swing calls for a continued activation of the hip flexors and plantar flexors in addition to the knee extensors. Each step causes a slight loss of mechanical energy because of the impact of the swinging leg and the ground. Gaining this energy back and continuing the gait cycle depends predominantly on work from the ankle and its
plantar flexors (Matjačić, 2009). Missing a single one of these muscle groups and more importantly, not being able to mimic the missing group with a prosthetic, makes completing a fluid gait cycle problematic.

**Prosthetic Design**

Prosthetics are artificial limbs used to replace anatomical structures in order to help regain some lost functions. With the example of a transfemoral amputee, also known as an above the knee amputee (AKA), the residual limb ends anywhere between the hip and the knee. The residual limb is most commonly fitted with an air tight socket which is connected to the rest of the prosthetic. Excessive rubbing or an uneven amount of pressure throughout the entire residual limb can cause serious skin irritation and pose additional safety concerns. Below the socket is the remaining prosthetic that emulates a knee joint, either with a simple hinge or an advanced microprocessor controlled knee. When specific muscle groups are lost due to amputation, the normal explicit muscle functions need to be mimicked by the prosthetic in order to achieve efficient mobility. Because of their lack of ankle and possibly knee, amputees experience difficulty trying to maintain stability in the anterior-posterior direction than any other direction (Lenka, 2010).

**Experimental Hypothesis & the Amputee Gait Cycle**

In this paper, 3D motion capture technology is used to quantitatively and qualitatively evaluate gait cycles in a digital experimental context. The null hypothesis is that in comparing a natural and a prosthetic compromised gait cycle, there is no significant change in gait mechanics. An alternative hypothesis is that the gait mechanics between natural and prosthetic compromised tests will be substantially altered and that the differential responses will provide insight on prosthetic design and therapy.
With a transfemoral prosthetic, initial contact begins with the knee hinge in a locked position. The hinge, as well as the bottom of the residual limb, will absorb the impact in the loading phase. During mid-stance, similar to normal walking, there is a passive rotation over the artificial foot. Push off, a crucial component to the gait cycle, is where an amputee must compromise due to the absence of plantar flexors. The majority of the power, consequently, must come from the hip. Without muscular control of a knee, power generated by the hip will have to come from a lateral swing motion, placing added stress on the hip abductors during the pre-swing phase (Matjačić, 2009). While it is less noticeable, gait in below the knee or trans-tibial prosthetics has comparable abnormalities as with transfemoral devices.

METHOD AND MATERIALS

Motion Capture

The hypothesis was evaluated by identifying the muscle systems that control gait locomotion, and corresponds those muscles with motion capture system sensors to digitally record the motion on test subjects. Following Tadano (2013), a subject wearing sensors on their iliac crest, anterior superior iliac spine, lateral epicondyle of femur, distal end of femur, medial condyle of tibia, tibial tuberosity, and the distal end of tibia was filmed using motion capture software MyCap Studio 2012 (Figure 1). MyCap Studio 2012 uses a dual camera overlap field of view hardware set-up with a sophisticated calibration to determine auto detected landmark motions in every frame of a movie shot at 25 frames per second.

The landmark configuration corresponds with muscle with careful consideration to their insertions and origins on the skeleton. Therefore, skeletal motion is used as a proxy for muscle system contractions. Calibrating the cameras and system requires moving a printed checkered
Figure 1. Normal and Prosthetic Compromised Data with MyCap Studio Sensors: Single frames taken from the same phases of the gait cycle for normal (A) and prosthetic compromised (B) data movies with the anatomical sensors or markers depicting corresponding anatomical markers. Note, software data automation does not recognize the same anatomical marker numerical order.
pattern within the cameras’ overlapping fields of view. Once the video was uploaded to the program, single frames of the checkered pattern at multiple positions within the frame are collected to derive a calibration for the 3D coordinate space within which our motion data is collected.

Field of view settings, distance of the calibration paper, and several other conditions required meticulous adjusting to yield a strong calibration. MyCap Studio 2012 collected data points within a Cartesian coordinate system (or X, Y, and Z axis system) for each sensor during the roughly 200 frames. The X and Z axis associated with horizontal and vertical movement respectively, and the Y axis represent depth that is parallel to the walking path. This quantitative data can be independently evaluated from the qualitative or visual evaluation of the raw film recording.

**The Pseudo-Leg**

Every individual has a unique gait influenced by age, weight, height, and fitness level. To control the number of variables evaluated in the experiment, the same test subject is recorded for a “natural” gait cycle and a prosthetic compromised gait cycle. To approximate the mechanical effect of a prosthetic limb affixed to a trans-femoral amputation a prosthetic, suitably named the “pseudo-leg”, was designed and built for able bodied test subjects. The leg consisted of an aluminum cane and hinge that allowed subjects to rest their bent knee into a metal U-shaped brace. The test subject’s leg was secured to the metal brace for stability and subjects were allowed to hold onto the sides of a treadmill for additional support while walking.
RESULTS

Video recordings of the normal and the prosthetic compromised gaits reveal a substantial qualitative difference in the gait cycles. There is a prominent asymmetry observed in the prosthetic compromised gait and more symmetrical cycle in the normal gait. Focusing on the right limb where the pseudo-leg was used, there is a visible lateral displacement of the knee in the swing phase of the prosthetic compromised gait compared to the normal gait swing phase. Unexpectedly, there is an additional slight lateral displacement in the hips observed in the prosthetic compromised gait that is not as apparent in the normal gait cycle. Therefore, the null hypothesis of an equivalent gait mechanics in normal and prosthetic compromised gaits is rejected based on the qualitative observation.

The magnitude and direction of the qualitative observations are further investigated using the quantitative data recorded from the motion capture system. Repeated complications with the MyCap Studio software have severely compromised the volume of data collected in the experiment. One difficulty in particular is the depth or Y-axis expressing irregular and error prone data (Figure 2). However, the MyCap Studio software was able to collect relatively consistent data in the horizontal and vertical directions (X-axis and Z-axis respectively). In addition, the automated tracking of the anatomical markers is inconsistent over the full 200 frame movie evaluation. As a result, evaluations of the full complement of 16 markers is not possible and quantitative evaluations is restricted to only a handful of markers of the knee and hip that are recorded for both normal and prosthetic compromised data runs over the full 200 frame recordings.

Two markers on the right distal femur are recorded from the medial and lateral sides, and the left iliac crest marker gives data on the normal and prosthetic compromised gaits.
Figure 2. Example XYZ Coordinate Data Output from MyCap Studio: MyCap Studio graphical output of the right tibia coordinate data through 200 frames for the normal (A) and prosthetic compromised (B) gaits for the same test subject. The lack of periodicity in the Y-Axis (yellow line) in the prosthetic data is indicative of an error in the automated tracking of the anatomical marker over the 200 frame data run.
A 4.6 to 4.4 centimeter (cm) range of lateral movement of the distal femur is observed in normal gait cycle measuring medial and lateral markers respectively (Table 1, and Figure 3). A 20.6 cm and 13.9 cm lateral displacement is observed in the prosthetic compromised gait cycle for medial and lateral markers (Figure 3). Finally, the left iliac crest marker in the normal gait records a 4.02 cm lateral displacement and a 17.158 cm lateral displacement in the prosthetic compromised gait cycles (Figure 4).

**DISCUSSION**

The qualitative results were analyzed for bilateral symmetry throughout the gait cycle as a guide. While an able bodied individual exudes a rather symmetrical walk, amputees do not. The muscle functions lost due to transfemoral amputation include the plantar flexors at the knee and ankle, dorsiflexion at the ankle, flexion and extension of the knee, and the digital flexors and extensors of the toes. These crucial muscles are necessary for completing the gait cycle and without adequately mimicking them with prosthetic design, existing muscles will need to take on more of the work load. Pin pointing the exact phases within the cycle amputees deviate from this normal bilateral symmetry is where therapists can identify the muscles of concern for overuse.

The loss of functionality and ultimately deviation from bilateral symmetry is especially noticeable between the push off and pre-swing phases. The hip flexors, typically having minimal participation within these phases, now have to compensate for the lack of plantar flexors such as the gastrocnemius. As seen in Figures 3 and 4, when the normal gait cycle was recorded, minimal range of motion was observed. This value is interpreted as representing the stability in the test subject’s gait. Contrary to the normal cycle, data recorded for simulated transfemoral
amputees with a prosthetic display a far greater range of motion, thus proving instability in the system. The graphs in Figures 3 and 4 are plotted in the horizontal and vertical axes, and the

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<th>Prosthetic Compromised</th>
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Table 1. Lateral Range of Motion Measured for Three Anatomical in Normal and Prosthetic Compromised Gaits: Horizontal (X-axis) data values for three MyCap Studio sensors showing their range of motion through 200 frames of data recording 4 gait cycles.
Figure 3. Increased Lateral Knee Movement of Prosthetic Compromised Gait: Isolated lateral and medial right knee markers for normal (red and purple circles) and prosthetic compromised (blue and green diamonds) depicting horizontal (X-axis) and vertical (Z-axis) displacement in centimeters (cm) during a 200 frame recording of 4 gait cycles. The prosthetic compromised gait records a three to five fold increase in the lateral motion of the right knee.
Figure 4. Increased Lateral Hip Movement of Prosthetic Compromised Gait: Isolated markers of the iliac crest for normal (red circles) and prosthetic compromised (blue diamonds) depicting horizontal (X-axis) and vertical (Z-axis) displacement in centimeters (cm) during a 200 frame recording of 4 gait cycles. The prosthetic compromised gait records a three to five fold increase in the lateral motion of the left hip as recorded by the iliac crest marker.
increased ranges of motion in the hip and knee are primarily in the horizontal axis with relatively minimal deviations throughout the gait cycle in the vertical direction. It can be concluded from this that deviation is due to lateral movement. In this experiment, subjects wearing the pseudo leg swung the amputated limp laterally on average three to five times (Table 1) further lateral than a healthy able bodied individual. This motion would result in additional stress on the adductor brevis and adductor longus muscles.

Some phases of the transfemoral simulated gait cycle did not show significant deviation in bilateral symmetry. These include mid-stance, which is primarily a passive movement, and initial contact. During initial contact, the body experiences an external movement from ground reaction forces in front of the hip which require hip flexors. Ankle dorsiflexors normally allow the foot to pivot anteriorly from the heel. Without a functioning hinged ankle joint in need of flexion, dorsiflexors are not needed.

In summary, this research finds that the loss of ankle function in a prosthetic leg propagates to alter the swing phase of the knee. This compensating motion that alters the path and range of lateral motion in the knee’s swing phase further propagates to the hips on the side with the prosthetic leg. Practice and experience walking with the prosthetic limb will likely reduce this laterally displaced gait of the knee and hips over time. However, physical therapy protocols for prosthetic limb use should expand to include core muscle exercises of the torso to improve control of the hips, and knee motion. Alternatively, future prosthetic design should focus on trying to improve the functional recovery of the ankle to avoid the knee and hip compensation in the swing phase of the gait cycle.
CONCLUSION

From the data collected in this study, the null hypothesis stating that there is no significant difference in gait mechanics between able bodied and transfemoral amputates can be rejected. Although sample sizes of the experiment were small, its results act as supporting evidence for the need for improved prosthetic design and physical therapy. Amputees are at an increased risk for unintentional falls due to their limited stability, their novel motion mechanics seen in the knee and hip to compensate for this loss of function make them highly susceptible to overuse injuries. While this cannot be completely rectified, steps can be made in therapy towards decreasing these risks by improving the swing phase of the prosthetic leg gait. This research proposes an intensive therapy approach in strengthening the torso to control hip motion and improved muscular endurance of the hip to control lateral knee motion. In response to gravitational pulls within the anterior-posterior plane, it is vital that the hip abductors and adductors are able to counteract it (Sethy, 2009). In addition to therapy completed immediately post amputation, the hip flexors and general core need to be the focal point of continuous therapy throughout an amputee's life.

The most difficult task in maintaining stability for transfemoral amputates is in the anterior-posterior plane. This can be credited to the missing muscles responsible for plantar flexion of the ankle and engineers need to be striving towards creating a suitable and affordable ankle joint that can recover a greater portion of this lost function.
References


