Kinetic and Kinematic Analysis of Fast and Slow Gait Speed

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Introduction

In order to move the human body forward, the body must coordinate the force production and angles of the body segments in the kinetic chain. When the body is in contact with the ground, there is a ground reaction force (GRF) acting equal and opposite to the weight of the body. Muscles can be used to manipulate the GRF in various ways. For example, when standing upright, muscles at the ankle are contracting to keep the location of the GRF close to the body's center of mass (in the x- and z-direction) to maintain stability (1). Similarly, the GRF can be shifted further away from the body's center of mass to propel the body forward. Force production does not occur only at the ankles, though. Forces are also produced by the muscles at the knees and hips when standing and walking. The forces produced at the various segments need to be coordinated in a way that is conducive to walking in an upright and stable position. Similarly, the angles and velocities of the joints also must be coordinated to move efficiently.

Motion capture and force plates can be used to assess the kinematic and kinetic parameters of gait (2). 3D motion capture is a technique where reflective markers are placed at various locations on the body and a camera records the positions of the markers over time. Kinematic data can then be derived from the positional data. Force plates are used to measure the magnitude and location of a force, and these two systems can then be used in concordance to provide kinetic and kinematic measurements over time. Although force plates can only provide information about the GRF over time, inverse dynamics can be used to analyze the forces, moments and power at various joints (3).

It is clear that the body must coordinate forces and angles of the kinetic chain to move forward, but what is less clear is exactly how this occurs. The purpose of this study is to gain a better understanding of how the body propels itself forward, and to assess which kinetic and kinematic parameters are altered to change gait speed.

Methods

One female subject (mass = 49.9 kg) was outfitted with reflective markers for motion capture (VICON) using a standard configuration. Markers were not placed at medial positions on the body. The volunteer performed one walking trial at normal speed and one at fast speed. A force plate (BERTEC) was used to capture the forces for a segment of the walking trials. Motion capture data was sampled at 100Hz, and force platform data was sampled at 1000Hz.

The force platform and motion capture data were time-locked. Since medial markers were not used, our analysis was confined to the sagittal plane. Filters were used to align the force platform coordinates with the global axis of the VICON system. Kinematic data was calculated, which included segment angular velocities and accelerations, joint angular velocities, center of mass and center of mass acceleration of each segment, mass and inertia of each segment, and stride length and cadence. Standard anthropometric tables were used to approximate center of mass, and segment mass and inertia (4). Stride length was calculated by finding the maximum horizontal displacement between the heel marker and hip joint marker (5). Cadence was calculated by using the time interval between successive heel strikes. Kinetic data was calculated, which included power and moments at each of the joints. The forces, moments, and power at joints were calculated using inverse dynamics equations.

Results

One gait cycle of the right foot was isolated for plotting. The cadence during the normal and fast trials were 111 and 132 (+18.9%) steps per minute, respectively. The stride lengths for the normal and fast walking trials were 0.783 and 2.12 (+170%) meters, respectively.

Figure 1

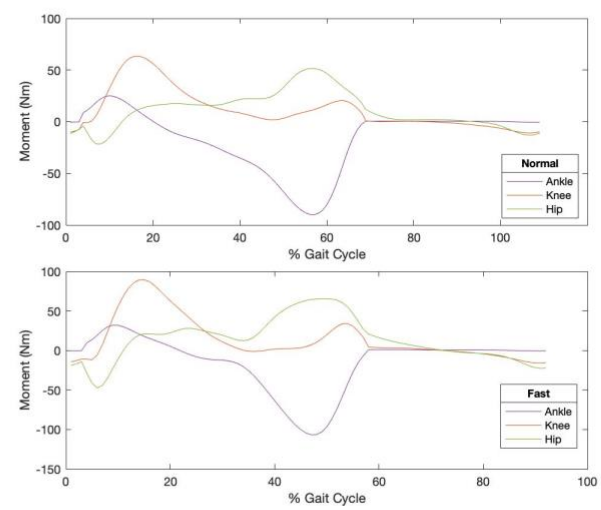
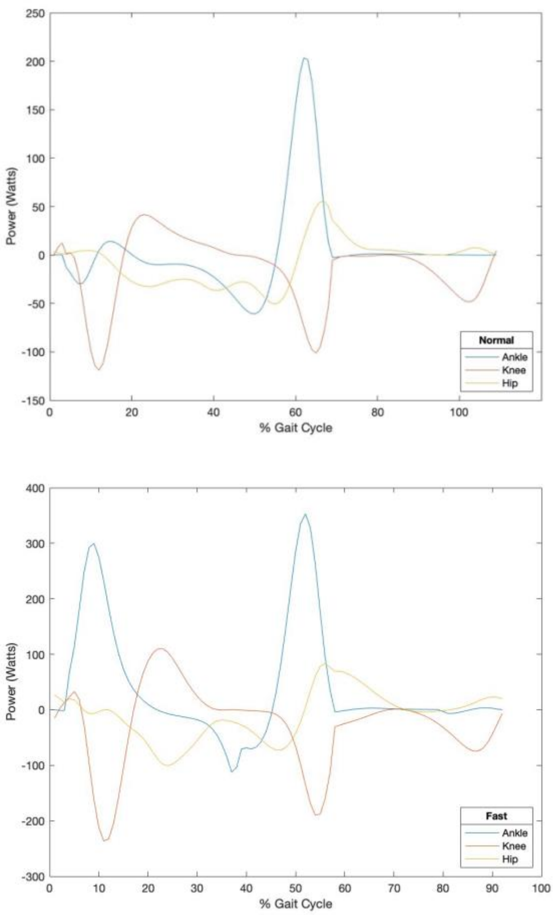


Figure 1 shows moments at the ankle, knee, and hip during normal (top) and fast (bottom) gait. Absolute peak torque at the ankle, knee and hip occurred at approximately 15%, 55%, 55% of gait cycle, respectively, during both normal and fast walking trials. The magnitude is greater for all muscles during the fast trial throughout the gait cycle.

Figures 2 & 3



Figures 2 and 3 show changes in power at the ankle, knee, and hip during one gait cycle during normal (top) and fast (bottom) walking trials. Absolute peak power occurred at approximately 60%, 15%, 60% for the ankle, knee, and hip, respectively, during both normal and fast walking trials. An ankle power output of ~300 watts occurred around 10% of the gait cycle during the fast trial, and this did not occur during the normal trial. The magnitude is greater for all muscles during the fast trial throughout the gait cycle.

Discussion

The plotting of moments during normal and fast gait speed suggests that the overall pattern of inverse dynamics remains the same. In both the normal and fast trials, the gait cycle is initiated by a small, negative hip moment with a positive ankle moment of roughly equal magnitude occurring simultaneously. Soon following is a positive moment at the knee, which was approximately 3 times the magnitude of the moment at the ankle in both the normal and fast trials. The knee moment was the global maximum for both the normal and fast trials. The moments of all three muscles reached a local minimum and maximum within the first 20% of the gait cycle for both normal and fast trials. Approximately 55% into the gait cycle the ankle, knee, and hip produce a second set of local minimum and maximum values for both trials. The ankle produced a global minimum for both the normal and fast trials between 45-55% of the gait cycle. At the same time, the hip and knee produced positive moments, with the knee moment being less than half the magnitude of the moment at the hip in both samples. The graphs suggest that, even when moving at different speeds, the pattern, relative magnitudes, and timing of moments occurring at the hip, knee and ankle remains relatively constant during gait.

There were many similarities between the moments of normal and fast gait, but what did change is the absolute magnitude of the moments produced by the muscles. The global maximum, which was produced by the knee, and global minimum, which was produced by the ankle, were roughly 50% and 11% greater, respectively, during the fast trial. The moments produced by the hip were also greater in the fast trial. It makes sense that greater torque was produced during the fast trial. Momentum is the product of mass and velocity, and momentum is equal to impulse, which is the product of force and time. In order to walk faster, that is, to move at a greater velocity, the product of force and time must increase (assuming mass remains constant). In the context of walking, the time component cannot increase because the goal is to walk more quickly; in order to walk more quickly, the feet must come in contact with the ground for a shorter period of time (6). Thus, force must increase to increase walking cadence. It is interesting to consider what happens to the length of the moment arm at the foot in this scenario. If the ankle is producing a greater amount of force, the center of pressure is likely shifting to a more positive x-direction, which would increase the moment arm between the center of mass of the center of pressure. If the moment arm increases, then the moment at the ankle will increase due to the combined effect of the increased force as well as the increase in the length of the moment arm. A systematic review by Fukuchi and colleagues found that hip flexion, knee extension, and ankle plantarflexion moments increased as gait speed increased in young adults (7).

The findings for lower extremity power between normal and fast gait were similar to that of the moments, in that the patterns, timing, and relative magnitudes were similar between trials but the absolute magnitudes were different. The local minimums of knee power were roughly doubled, the hip global maximum was roughly doubled, and the ankle local maximums were roughly 50% greater in magnitude in the fast trial. Aside from differences in magnitude, the patterns were very similar between trials except for one notable exception. A local maximum for the ankle (~300 watts) was observed within the first 10% of the gait cycle for the fast trial, and this did not occur in the normal trial. One possible explanation is that, since the body was moving more quickly just prior to the heel strike phase, a greater power would've been required to overcome those forces and propel the body forward. Why a similar pattern didn't occur for the ankle moment during the fast trial is unclear. Since power has a time component, it may be the case that the heel was in contact for a much shorter period of time compared to the normal trial, which would lead to an increase in ankle power but not necessarilty ankle torque.

Cadence was faster and stride length was greater for the fast trial. It appears that stride length and cadence are two ways the body increases gait speed, which is an agreement with the findings from a meta-analysis that investigated the effects of walking speed on gait biomechanics in healthy young adults (7). Stride length increased by 170% compared to normal gait. If the body is applying a greater force to walk faster, then it would make sense that the next step taken is a greater distance away because the forces are propelling the body forward at a greater velocity, and more distance would be covered during each step.

There are several limitations to our findings. Our kinematic and kinetic analysis was confined to the lower extremities, and it is unknown what effect the upper body kinematics would've had on our outcomes. For example, it is not known if the trunk position changed between normal and fast gait. Changes in trunk position can change the position of the center of mass, which could change the amount of torque needed to propel the body forward. Another limitation to this study is that it relies on the accuracy of marker positional data. Small errors in marker placement or marker readings will lead to errors in moments, positional data, and all subsequent positional data calculations. This limitation was partially controlled for by confining the analysis to the sagittal plane. Because movements are much larger in the sagittal plane, skin movement will not cause as large of an error in this dimension compared to the transverse and frontal planes, where movements are smaller and skin movement errors will comprise a larger proportion of the variance. Another limitation is the assumption of a rigid body for kinetic calculations. It is assumed that all forces are transferred from one end of the segment to the next, but this is a simplification; deformations of soft tissues will lead to less than all of the forces being transferred (8), which will affect moment and power calculations. Lastly, a very small amount of force data was collected, and it was only collected for the right foot. If the volunteer has asymmetries in their gait patterns, this may have led to values that are not representative of the individual's overall gait. Stride length and cadence were calculated by using only one distance and time measure between the right heel contact phase. These values may not be perfectly representative of the average stride length and cadence of normal and fast gait.

Kinetic and kinematic parameters are manipulated to propel the body forward. How exactly the body does this is unclear, and the purpose of this study was to investigate the ways in which these parameters can be altered to move the body forward at different speeds. It appears that the kinetic patterns remain relatively constant when moving at normal and fast speeds, but the magnitude of the torques and power at the joints increases. This is likely due in part to the increased force production at the various segments in the body, as well as an increase in moment arm length between the center of pressure and center of mass of the foot. The increase in power for fast walk can be explained by an increase in force production combined with a decrease in foot contact time. Additionally, stride length and cadence appear to be two mechanisms the body uses to increase gait speed.

References

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