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Ankle torque control that shifts the center of pressure from heel to toe contributes non-zero sagittal plane angular momentum during human walking

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ABSTRACT

A principle objective of human walking is controlling angular motion of the body as a whole to remain upright. The force of the ground on each foot (*F*) reflects that control, and recent studies show that in the sagittal plane *F* exhibits a specific coordination between *F* direction and center-of-pressure (CP) that is conducive to remaining upright. Typical walking involves the CP shifting relative to the body due to two factors: posterior motion of the foot with respect to the hip (stepping) and motion of the CP relative to the foot (foot roll-over). Recent research has also shown how adjusting ankle torque alone to shift CP relative to the foot systematically alters the direction of *F*, and thus, could play a key role in upright posture and the *F* measured during walking. This study explores how the CP shifts due to stepping and foot roll-over contribute to the observed *F* and its role in maintaining upright posture. Experimental walking kinetics and kinematics were combined with a mechanical model of the human to show that variation in *F* that was not attributable to foot roll-over had systematic correlation between direction and CP that could be described by an intersection point located near the center-of-mass. The findings characterize a component of walking motor control, describe how typical foot roll-over contributes to postural control, and provide a rationale for the increased fall risk observed in individuals with atypical ankle muscle function.

1. Introduction

From a whole-body perspective, the mechanical objective of walking is to use the legs to move the body though space while preserving upright posture. That postural task necessitates regulation of whole-body angular position, which has received relatively little attention beyond observational studies of angular motion during human walking (Popovic et al., 2004; Bennett et al., 2010; Herr and Popovic, 2008). Irregular control of angular body motion is related to an increased risk of falling (Nott et al., 2014; Simoneau and Krebs, 2000), but the mechanism by which humans regulate angular motion while walking is poorly understood. One outcome of neuromuscular control is ankle torque, which is important for maintaining appropriate angular body motion (Pijnappels et al., 2005). Studies of subjects with a history of falling have shown atypical ankle muscle function when compared to non-fallers (LaRoche et al., 2010; Pijnappels et al., 2005; Simoneau and Krebs, 2000; Skelton et al., 1999, 2002). Application of this evidence to improve fall prevention strategies and gait rehabilitation requires a more precise understanding of how ankle function combines with the other neuromechanical interactions to produce typical walking behavior. Therefore, this paper presents a model of a potential mechanism by which neuromuscular control of the hip, knee and ankle torques and segmental mechanics enable control of sagittal-plane angular posture during human walking.

Previous walking studies of sagittal-plane whole-body angular position have focused on characterizing the force of the ground on the foot (*F*) through stance (Maus et al., 2010; Gruben and Boehm, 2012b). The effect of neuromuscular control acting on the body (a mechanical linkage) is reflected in both the motion of the body and the net force acting on the body. The previous work chose to quantify *F* and investigated how properties of *F* such as location of application (center of pressure, CP) and direction (Fig. 1) co-vary during walking to preserve upright posture.

These CP and direction properties of *F* are key features of gait to observe when focusing on how people prevent tipping over, because they determine the torque of *F* about the center-of-mass (CM) during unaided walking. *F* is the only force acting on the body during unaided walking that can produce torque about the

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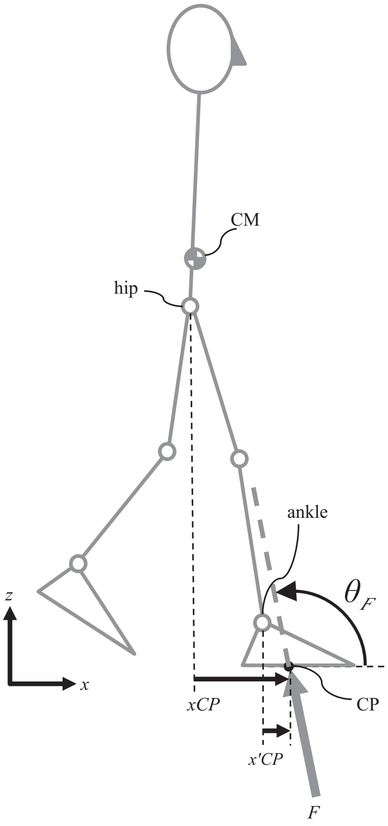

Fig. 1. Factors contributing to xCP location. The location of the CP of F expressed relative to a vertical line through the hip (xCP) depends on (1) the foot location relative to the hip and (2) the location of CP relative to the foot ($x'CP$). A vertical line through the ankle joint was used to reference $x'CP$. The direction of F (θ_F) was expressed relative to the x axis.

CM, and thus, is directly related to the whole-body angular acceleration that determines whether someone tips over. Characterization of F through the gait cycle with respect to the CM, then, is sufficient on a whole-body level to characterize how the system prevents tipping over. Discussion in previous reports of walking studies has been limited to this characterization of the overall nervous system output provided by observing the F relative to the CM. Those studies have shown that the F direction and location (CP) co-vary in a pattern that is common across individuals (Maus et al., 2010; Gruben and Boehm, 2012b). That typical nervous system control results in a systematic pattern of non-zero angular acceleration alternating in sign such that it averages zero (Herr and Popovic, 2008). Through the single limb stance phase of the gait cycle, the CP varies from anterior to posterior relative to the vertical line through the CM. Through the same time, the direction of F begins angled backward (the F has a horizontal component directed posteriorly) with a line-of-action passing anterior to the CM, is then vertical, and finally is angled forward with a line-of-action passing posterior to the CM (Fig. 2) (Maus et al., 2010; Gruben and Boehm, 2012b). That characterization of the typical pattern of F variation was shown to promote upright posture, because it produced a torque in the direction needed to pitch the body toward upright throughout most of the gait cycle.

The present study initiated a more detailed investigation of F during walking by focusing on the contributions to CP position. During single limb support, the motion of the CP relative to a vertical line through the hip can be dissected into two factors; (1) posterior motion of the stance foot relative to the hip (stepping), and (2) motion of the CP with respect to the foot (foot roll-over) (Fig. 1). The foot roll-over factor is typically described as heel-to-toe, as the foot initially contacts the floor with the CP near the heel and leaves the ground with the CP near the toe. Heel-to-toe and alternate CP patterns such as those used in toe walking (Crenna et al., 2005) and speed change (Orendurff et al., 2008)

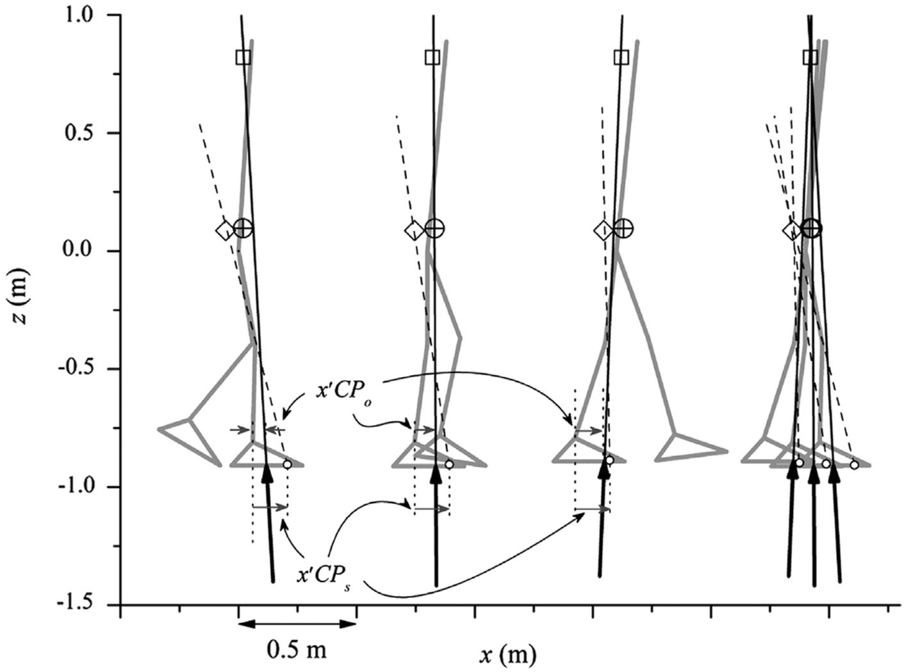

Fig. 2. Characteristics of the *observed* F and of the *modified* F that is positioned at a specified $x'CP_s$. In the three left figures, different instants from a representative walking cycle are shown with both the *observed* F (bold line-of-action and vector) and line-of-action of the *modified* F (dashed line) that results from selecting ankle torque to position the CP at the specified point ($x'CP_s$, open circle). The *observed* F lines-of-action pass from the observed CP ($x'CP_o$) to near a DP (square that is fixed with respect to the hip across postures). The *modified* F lines-of-action pass near the point *xi* (diamond that is fixed with respect to the hip across postures) after shifting the CP to the same specified point on the foot ($x'CP_s$, open circle) but keeping the same body segment positions, velocities, and knee and hip torques as observed at each of the three instants. The right panel shows the three instants superimposed with the hip at the origin of the reference frame to illustrate the proximity of the *observed* F lines-of-action to the DP and the proximity of the *modified* F lines-of-action to *xi*.

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have been attributed to varied control of ankle torque. Consequently, this dissection is valuable for providing insight on control of upright posture, because it separates the role of ankle torque modulation and its effect on *F* from other contributions (hip and knee torques) to walking behavior.

The dissection is enabled by previous characterization of the CP and *F* direction co-variation strictly due to ankle torque modulation (Gruben and Boehm, 2012a). Consider that at any instant in the gait cycle, the kinematic state of the body can be defined by the limb segment positions and velocities. It was shown that for any specific kinematic state, any distinct ankle torque value produces a unique *F* with its own corresponding CP and direction (Gruben and Boehm, 2012a). That relation between ankle torque and *F* is due to the mechanics of muscles acting on the body and interactions of segments modeled as a linkage of rigid bodies. This approach enables quantification of the foot roll-over contribution to observed walking behavior.

Therefore, to better characterize a component of neuromuscular control, this study employed modeling and empirical data to describe the observed relationship between the *xCP* and *F* direction in human walking (Maus et al., 2010; Gruben and Boehm, 2012b) separated into terms of ankle torque control that shifts the CP with respect to the foot (foot roll-over) and hip and knee torque control that shifts the CP by regulating motion of the stance foot relative to the hip (stepping). Both factors affect *F* and thus, aid in understanding how humans control angular posture.

2. Methods

To quantitatively separate the stepping and roll-over aspects of walking, the human body was modeled using a seven-segment linkage articulated to move in the sagittal plane (see Section 2.2 below). Force and kinematic data were collected from human subjects in steady-state walking on a custom force treadmill for 30 s (specific methods in (Gruben and Boehm, 2012b)). The treadmill design placed force plates in direct contact with each foot to provide the CP and *F* measurement precision needed for this analysis. Hip, knee, and ankle joint torques were calculated using inverse dynamics.

The stepping and foot roll-over factors in the walking data were treated as follows: (1) Stepping was defined to exclude change in CP relative to the foot, so an arbitrary referent location on the foot was specified as a fixed CP location ($x'CP_s$, Fig. 2). To analyze an instant in single limb stance, the ankle torque consistent with the $x'CP$ at the fixed $x'CP_s$ rather than at the observed location was determined (Appendix A). Forward dynamics with the new ankle torque value fully specified the instantaneous *modified F* at the fixed $x'CP_s$, which was generally distinct from that observed. (2) The foot roll-over factor consisted of the difference in *F* direction and CP between the *observed F* and *modified F*. That factor was due to the difference between the observed ankle torque and that needed to set the CP at the fixed $x'CP_s$. Note that a difference in magnitude also generally existed between the *observed F* and *modified F* but was not included in the scope of this study due to its much smaller relative contribution to the change in torque *F* produced about the CM. The sensitivity of the torque *F* produced about the CM to the direction, magnitude, and location of *F* was quantified to substantiate this simplification (Appendix B).

By evaluating multiple postures in a gait cycle in this manner for the same $x'CP_s$, this analysis can be conceived of as exposing the stepping factor effect of CP shift on *F* direction by removing the effect of the heel-to-toe CP shift from the *observed F* direction. The resulting *modified F* characterized the neural control of hip and knee torques, expressed graphically as the pattern of *modified F* vectors across multiple postures. This was accomplished by applying the procedure described in the previous paragraph at 10% time increments (11 points) spanning single limb support in the gait cycle to produce a set of 11 *modified F* vectors all with the CP fixed at the same specified location relative to the foot (Fig. 2). Ten percent time increments were chosen to sufficiently represent the dynamics of the signals. When expressed in a hip-centered coordinate frame, those *modified F* vectors had lines-of-action that passed near a point fixed on the body relative to the hip. That point was called *xi* (Fig. 2), and was located in the sagittal plane at (xi_x, xi_z) with respect to the hip.

It is important to recognize that the choice of $x'CP_s$ is simply a reference frame choice that allows for a single point, *xi*, to graphically represent the hip and knee torque control present during walking. While hip and knee torques have been observed in previous studies, representing their coordination in *F* vector space with the effect of ankle torque removed allows for characterization of how hip, knee, and ankle torques contribute to whole-body angular motion during walking by exposing the torque they induce about the CM via *F*.

The location of *xi* was calculated by minimizing the sum of squared perpendicular distances between the 11 *modified F* vector lines-of-action and the fixed point *xi*. To characterize the effect of the $x'CP_s$ choice on *xi* location, the above analysis was carried out on each subject for 21 distinct specified CP locations ($x'CP_s$) on the foot that were spaced at 5% increments from heel-to-toe of the shoe. For each $x'CP_s$, the proximity of the *modified F* vector lines-of-action to *xi* across the gait cycle was quantified with the RMS moment arm from the *modified F* to *xi*. The proportion of the variability in the *modified F* direction that was predicted by *xi* was quantified with the coefficient of determination (R^2, (Weisberg, 1985)) (Appendix C.i).

Previous work has shown that the co-variation of *observed F* direction and CP during walking may be characterized by the geometrical pattern that is the *F* vector lines-of-action passing near a point fixed with respect to the body, called the divergent point (DP, (Gruben and Boehm, 2012b), Fig. 2)). In the present study, the location of the DP was also determined by minimizing the sum of squared perpendicular distances between the *observed F* vector lines-of-action and the DP. The proximity of the *observed F* lines-of-action to the DP was quantified with the RMS moment arm from the *observed F* to the DP. The proportion of the variability in the *observed F* direction that was predicted by the DP was quantified with the coefficient of determination (R^2) (Appendix C.ii).

That previous work also showed that the location of the DP relative to the CM has implications for understanding how humans control whole-body angular motion during walking. Therefore, the proximity of the *xi* values to the CM location (estimation methods in Appendix D) was quantified by calculating the moment arms of the *modified F* vectors about the CM. For each walking cycle, the interpolated $x'CP_s$ location that yielded the minimum RMS moment arm to the CM was identified along with the corresponding *xi* location.

A location for the DP and the proximity and variability measures (RMS and R^2) were calculated for each walking cycle of each subject (13–18 cycles per subject × 11 subjects = 163 cycles). A location for *xi* and proximity and variability measures (RMS and R^2) were calculated for each walking cycle of each subject at each of the 21 $x'CP_s$ locations across the sole of the foot (13–18 cycles per subject × 11 subjects × 21 $x'CP_s$ locations = 3423 *xi* locations). Comparisons used Student's *t*-test with a significance threshold of 0.05.

2.1. Subjects

The subjects provided informed consent to the protocol approved by a University of Wisconsin Institutional Review Board. The 11 volunteer human subjects were aged 32 ± 9 years (mean ± standard deviation), height 1.73 ± 0.11 m, mass 73.3 ± 15.6 kg, and included 5 females. A representative subject was chosen for graphical illustrative purposes as the subject that was closest to the mean on multiple analyzed variables.

2.2. Model

The model consisted of a single segment for the head arms and trunk and two separate legs, each having a separate segment for the thigh, lower leg, and foot. The foot in single-leg stance was modeled as having zero acceleration and remained in full sole contact with the floor. Anatomical landmarks were used to measure segment lengths. For the inverse and forward dynamics modeling, segment CM locations, masses, and moments of inertia were calculated from anatomical landmarks and anthropometrics scaling factors (de Leva, 1996). This method is distinct from the whole-body CM location method that used subject-optimized parameters (Appendix D). A hip-centered, vertically-oriented reference frame was used to express all results with respect to the body (Gruben and Boehm, 2012b).

3. Results

For each walking cycle, the relationship between CP and *F* direction across time for both the *observed F* and the *modified F* obtained by removing the foot roll-over element were well-described as *F* lines-of-action that passed near points fixed with respect to the hip (Fig. 2). The moment arms of the *observed F* about the DP had an RMS value of 0.0036 m across all cycles and subjects. The DP described nearly all the variation in direction of the observed *F* (mean $R^2 = 0.98$). For a given gait cycle there existed a location of $x'CP_s$ that yielded *modified F* vectors that produced the smallest RMS moment arm about the CM. The moment arms of the *modified F* about CM produced at that $x'CPs$ had an RMS value of 0.0082 m across all subjects and gait cycles. The moment arms of the *modified F* about *xi* produced at that $x'CP_s$ had an RMS value of 0.0025 m across all subjects and gait cycles. The RMS *modified F* moment arms about *xi* indicate how well the hip and knee torque coordination across the gait cycle can be described by the *modified*

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F intersecting at a point (*xi*). This *xi* described nearly all of the variation in the direction of the *modified F* (mean $R^2 = 0.995$).

The locations of the *xi* were in a region distinct from the DP. The DP (*observed F*) clustered near the shoulders. For each gait cycle, the locations of *xi* (*modified F*) for the range of $x'CP_s$ spanning from the heel to the toe lay on a curve (Figs. 3 and 4). Across cycles within a subject, those curves typically had a similar shape and, thus, were summarized with a mean curve (Fig. 4). Those mean *xi* curves passed through a region adjacent to the CM (Fig. 4). The *xi* curves were typically sloped such that an $x'CP_s$ in the heel region yielded *xi* locations anterior and inferior to the CM while an $x'CP_s$ in the toe region yielded *xi* locations posterior and superior to the CM (Fig. 3).

The mean location of *xi* at the $x'CP_s$ which produced *modified F* with the smallest moment arms about the CM for each subject was located in a region near the CM (Figs. 3 and 4). The mean location of these *xi* was at (-0.003, 0.036) of CM height relative to the CM ($p < 0.0001$), which was significantly different from the location of the CM. The mean location of this $x'CP_s$ was 60% (standard deviation = 8%) of shoe length from the posterior aspect of the shoe. The first metatarsal head was located at 65% of shoe length from the posterior aspect of the shoe. The moment arms of the *modified F* about the CM for this *xi* had and RMS value of 0.008 m across all subjects and gait cycles.

As reported previously and confirmed here, the *F* measured during human walking was characterized by co-variation of *observed F* direction and the *xCP* such that the *observed F* lines-of-action passed near a point (divergent point, DP) fixed in a hip reference frame (Fig. 2) (Maus et al., 2010; Gruben and Boehm, 2012b). The mean location of the DP across all cycles and subjects was located an average of (-0.006, 0.594) of CM height relative to the CM ($p = 0.001$ and $p < 0.0001$, respectively) (Fig. 4).

4. Discussion

The purpose of this study was to better understand the organization of the neuromuscular control and its interaction with body mechanics that results in the observed pattern of angular acceleration previously shown to be favorable for maintaining upright posture during human walking. Most humans walk with a typical pattern of whole-body angular acceleration and force of the ground on the feet despite the latitude in neuromuscular control options which can still produce successful steady-state walking (Maus et al., 2010; Gruben and Boehm, 2012b; Herr and Popovic, 2008). Atypical regulation of angular motion during walking has been shown to correlate with risk for falling (Nott et al., 2014; Simoneau and Krebs, 2000), and thus, a model of how humans regulate that angular motion furthers understanding of neural control structure and can aid fall prevention research.

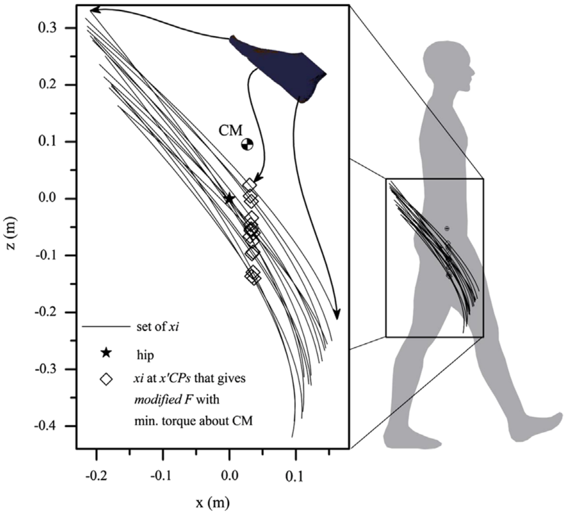

Fig. 3. Variation in *xi* location with choice of specified $x'CP$ location. The location of *xi* varied with choice of $x'CP_s$. For a single representative subject, a curve represents the locations of *xi* across all possible $x'CP_s$ locations (heel-to-toe) for a given walking cycle. All walking cycles for the representative subject are shown. When $x'CP_s$ was set at the heel, *xi* was located anterior and inferior to the hip (0,0). Conversely when $x'CP_s$ was set at the toe, *xi* was located posterior and superior to the hip. The $x'CP_s$ that produced *modified F* vectors that passed most closely to the CM yielded *xi* locations inferior to the CM for this subject (diamonds). The plot is superimposed on a body outline to provide an approximate scale reference.

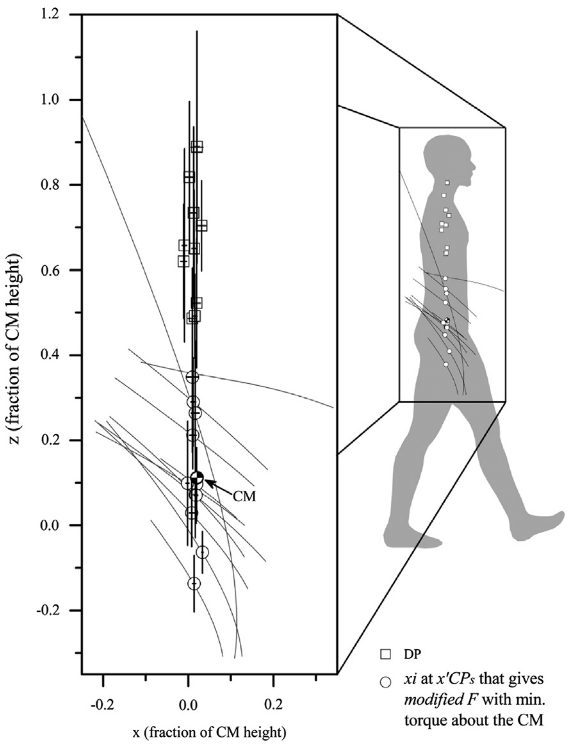

Fig. 4. Location of *xi* and DP. A curve for each subject shows that subject's mean (across walking cycles) location of *xi* across $x'CP_s$ choice within the foot (hip at (0,0)). Across the 11 subjects, the *xi* locations at the $x'CP_s$ that predicted modified *F* with the smallest RMS moment arms to the CM are distributed superior and inferior to the CM (circles, ±1 standard deviation whiskers). Four subjects had mean *xi* locations slightly superior to the others. The mean CM location across the 11 subjects and across the gait cycles was anterior and superior to the hip (0,0). The locations of the divergent point (DP) for each subject (squares, ±1 standard deviation whiskers) were located superior to the *xi* and nearly centered vertically above the CM. The plot is superimposed on a body outline to provide an approximate scale reference.

Because the force of the ground on the foot (*F*) results primarily from neuromuscular activation and is directly related to whole-body angular acceleration, its properties provide information about the strategy that the nervous system uses to control upright posture. This study built on previous work (Gruben and Boehm, 2012b) that described how two of those *F* properties, the location of application (CP) and direction of *F*, co-vary in typical walking in a manner such that the body angularly accelerates toward upright through most of the gait cycle (Appendix E). In the present work, observed walking behavior was separated into CP and *F* direction changes due to foot roll-over and due to stepping. This dissection provided a possible explanation for what ankle torque control (foot roll-over) and hip and knee torque control (stepping) each contribute to the *xCP* and *F* relation (Appendix F). It appears that the stepping contribution alone may be adequate for walking upright but does not produce the beneficial angular acceleration towards upright

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through most of stance noted in overall walking behavior (Appendix E). The stepping element, rather, is shown here to be associated with *modified F* lines-of-action that pass closer to the CM (*xi*) and thus, is associated with smaller whole-body acceleration than observed in typical walking humans (Fig. 2). As suggested by the present study, a reason humans may choose to coordinate ankle torque with stepping (Appendix G) to cause a heel-to-toe CP shift is to exploit linkage mechanics and thus, alter *F* direction away from *xi* and toward the DP above CM, where *F* can produce a righting torque (Appendix E).

The difference between the DP and *xi* is particularly interesting for an $x'CP_s$ in the forefoot region. Across subjects, *xi* locations calculated from $x'CP_s$ at a mean of 60% foot length from the heel were clustered near the CM. This suggests the possibility of a preferred ankle torque configuration used by the nervous system to direct *F* from an $x'CP$ under the forefoot to near the CM, producing little angular acceleration. By not inducing angular acceleration, this strategy would be adequate for walking but would require corrective control to adjust the line-of-action of *F* if angular motion accumulates due to errors. Apart from error correction, use of the *xi* strategy without foot roll would only require hip and knee torque coordination in order to keep *F* directed through the CM and could utilize a simpler ankle torque control that keeps the CP on the forefoot. This simplified control may explain why the most common types of walking with atypical $x'CP$ excursion involve constraining the $x'CP$ to the forefoot as seen in walking development (Hallemans et al., 2006), habitual toe-walking (Kerrigan et al., 2000), cerebral palsy (Hsue et al., 2009), and stroke (Kinsella and Moran, 2008).

The $x'CP$ excursion of those atypical gaits is similar to digitigrade and unguligrade walking. Those walking styles have certain advantages (Adamczyk and Kuo, 2013) and occur more frequently in nature than the plantigrade walking observed in humans (Alexander, 1990), so the utility of this heel-to-toe configuration in humans is unresolved. Previous work has shown the heel-to-toe pattern to be energetically favorable over walking with a CP fixed on the foot due to reduced step-to-step transition energy cost (Adamczyk et al., 2006). It is possible for humans to walk without the plantigrade foot posture that allows for foot roll-over and to instead use some other pattern of lower limb torques to yield the DP behavior, as is observed in canine and galline digitigrade gait (Maus et al., 2010). This study shows, however, that humans typically appear to produce a supra-CM DP by employing an approximately CM-centric hip and knee torque control coordinated with an ankle torque that produces heel-to-toe foot roll-over. The result is an energetically favorable (Adamczyk et al., 2006; Usherwood et al., 2012) means of producing the DP behavior that also promotes upright posture. Under certain conditions, however, humans walk with atypical CP excursion. These include developing walking (Hallemans et al., 2006), habitual toe-walking (Kerrigan et al., 2000), cerebral palsy (Hsue et al., 2009), and stroke (Kinsella and Moran, 2008). Thus, the control strategies those populations use to control angular motion may lack energetic advantages and the stability advantages of typical human walking described here.

In addition, the CM-centric hip and knee torque control (*xi*) may be useful for tasks other than walking. When an $x'CP$ under the forefoot is present, *xi* shows that the force direction would be approximately CM-centric. That would produce force that does not induce angular acceleration, a critical characteristic for maintaining upright posture when a person suddenly increases the magnitude of *F* to jump (Mathiyakom et al., 2006). Pushing with the forefoot also maximizes contact time by effectively lengthening the leg. While *xi* was not located at the CM in all subjects, its proximity to the CM supports the idea that a CM-centric control, characterized by *xi*, is shared across tasks (d'Avella and Bizzi, 2005; Chvatal and Ting, 2013).

To avoid misinterpretation, it should be recognized that the *modified F* vectors at a fixed $x'CP_s$ used to locate *xi* in this paper are distinct from that which would be observed in a subject physically walking with a constrained $x'CP$ but otherwise typical kinematics. The instantaneous decomposition of *F* used in this study yields *modified F* that are *not* consistent with the observed body state at the following instant, because those *modified F* are only a partial contribution to the *observed F*.

This study showed that humans may couple ankle torque-induced foot roll-over with an approximately CM-centric knee and hip torque strategy to produce the *F* that maintains upright posture during walking. This strategy is energetically favorable (Adamczyk et al., 2006) and appears to be well-suited for upright posture preservation during jumping. The beneficial pattern of the righting torque that *F* produces about the CM appears to be the result of ankle torque control producing foot roll-over coordinated with stepping. This provides a possible explanation for why many studies find that fall-prone individuals exhibit atypical ankle torque patterns during walking (Pijnappels et al., 2005; LaRoche et al., 2010; Skelton et al., 1999, 2002; Simoneau and Krebs, 2000). Atypical ankle torque may not produce the characteristic DP behavior and the accompanying stability advantages. This understanding may be useful in designing gait restoration and fall prevention therapies. Development of mechanical devices that walk more like humans or aid human walking may also be facilitated.

Conflict of interest statement

The authors report no conflict of interest.

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Appendices: Supporting information

Supplementary data associated with this article can be found in the online version at http://dx.doi.org/10.1016/j.jbiomech.2014.01.034.

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Ankle torque control that shifts the center of pressure from heel to toe contributes non-zero sagittal plane angular momentum during human walking

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Appendix A: *Methods for Determining Modified F*

The linear relationships reported in a previous study (Gruben and Boehm, 2012a) among ankle torque, the horizontal component of F (F_x), the vertical component of F (F_z), and the product of $x'CP$ and F_z meant that there also existed a linear relationship between $x'CP$ and the ratio F_x/F_z. For the observed body state at that instant, forward dynamics was used to calculate F_x/F_z and $x'CP$ at two arbitrary ankle torque values. A fit to the linear relationship between F_x/F_z and $x'CP$ allowed for extrapolation to find the F_x/F_z at the specified $x'CP$ ($x'CP_s$).

Appendix B: *Contribution of F Magnitude*

An additional route for further investigation would be to quantify the change in F magnitude associated with $x'CP$ shift and its implications for controlling angular body motion. Using the model in this study, the normal and friction components of F each have a linear relationship with ankle torque at a given body state. Thus, relocating $x'CP$ via ankle torque would not only alter the torque potential of F about the CM due to moment arm change, but also F magnitude change. The effect of F magnitude change was not included in the present study, because its influence on the torque produced about the CM was relatively much smaller than that of the corresponding F direction and CP changes. A sensitivity analysis of the torque F produced about the CM to F magnitude showed a contribution of only 2.8% ± 2.9% (mean ± SD over all subject, all cycles) of that due to the combined effect of magnitude, direction, and CP location. F direction accounted for 65.6% and CP contributed 31.6%.

Appendix C: *Coefficient of Determination Estimation*

i) $R^2 = 1 - [\Sigma(\theta_{mod}\text{-}\theta_{xi})^2 / \Sigma(\theta_{mod}\text{-mean } \theta_{mod})^2]$
where θ_{mod} is the direction of the *modified F* and θ_{xi} is the direction of the line from $x'CP_s$ to *xi*.

ii) $R^2 = 1 - [\Sigma(\theta_{obs}\text{-}\theta_{DP})^2 / \Sigma(\theta_{obs}\text{-mean } \theta_{obs})^2]$
where θ_{obs} is the direction of the *observed F* and θ_{DP} is the direction of the line from $x'CP_o$ to the *DP*.

Appendix D: *Methods for Center of Mass Measurement*

The whole body CM location used for those moment arm calculations was determined using empirical measurements of body mass distribution for each subject. Vertical CM location was determined by first measuring the distance from the foot sole to the CM while the subject lay on a balance board. It was then assumed that the CM height varied during walking by the same amount as the pelvis, of which motion was measured. Due to the nearly vertical nature of the *F* vectors, moment arm calculations were particularly sensitive to anterior-posterior location of the CM. To obtain precise estimates of whole body CM along that horizontal axis, the subjects stood stationary on the treadmill in various poses spanning the kinematic space of walking. From measurements of body segment positions and knowing that the CM was located vertically above the measured CP, mass distribution parameters were estimated using an optimization algorithm. Those optimized body segment parameters predicted the anterior-posterior location of the CM for the optimized data with an error of 0.002 m (RMS across all 11 subjects).

Appendix E: *Righting Torque Character of DP Behavior and Perturbation Implications*

The overall divergent point (DP) behavior of the observed *F* emerges from the systematic muscle activation choices that the nervous system makes and the mechanical interactions of muscles with body segments. The righting characteristic of this DP behavior is present for an average of 87% of the time spent in single limb support for the subjects in this study. That property follows from a DP located above the CM, which indicates *F* being directed anterior to the CM when the body is pitched forward and F being directed posterior to the CM when the body is pitched backward. This results from the interaction of body mechanics with the coordination of the typically observed anterior CP shift with anterior body tilt and posterior CP shift with posterior body tilt. This coordination of CP shift with body tilt appears to be a consistent behavior during walking and would be expected to continue for small variations, deviations, and errors, though further study is needed to address the CP shift that may occur for larger body tilt perturbations. In typical walking though, the higher the DP is above the CM, the larger moment arm *F* has to produce a righting torque (this assumes that the *F* magnitude during single limb support cannot decrease significantly due to the requirement of supporting body weight).

Appendix F: *Lower Limb Joint Torque Contributions to $x'CP$*

Forward dynamic modeling showed that changes in ankle torque shifted the CP 2.4 times more than the same change in hip torque and 2.9 times more than the same change in knee torque (averages across all cycles and subjects). These factors support the choice of ankle torque modulation as a means to quantify the roles of stepping and foot-roll in CP shift during walking.

Appendix G: *Presence of Independent Ankle Torque Control*

It should be recognized that the interpretation of the study presented here relies on a plausible control strategy that implements ankle torque control independent of control of hip and knee torques. Such a control is present when humans change walking speed (Orendurff et al., 2008). While one particular way in which the hip, knee, and ankle torques may be coupled is due to the mechanical action of biarticular muscles, the presence of monoarticular muscles spanning each joint provides the nervous system with the capacity to independently control the torques at the hip, knee, and ankle.

Independent control of ankle torque was also supported by an additional observation of producing *modified F* vectors using the procedure described in this paper but instead modulating hip or knee torque instead of ankle torque. The *modified F* vectors produced by modulating hip or knee torque to shift the $x'CP$ to a specified location were not well characterized with a fixed intersection point, nor were results as similar across subjects as was the case for ankle torque modulation presented here. This lack of consistency and pattern of systematic covariation from hip or knee torque analysis point toward significance in the patterns found in the ankle torque analysis described here.