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Original Article

An Evaluation of the Correlation between the Free Moments Applied on the Lower Extremity and the Knee Extensor Mechanism Force in Pronated Foot Subjects during the Stance Phase of Gait

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ABSTRACT

Background: Due to the rotatory nature of the excessive subtalar pronation and the possible impairment of the tibial rotation-knee flexion mechanism, changes of the free moment (FM) and changes of the extensor mechanism force are expected in hyper-pronated foot subjects. The purpose of this study was to evaluate the correlation between the FM applied on the lower extremity and the knee extensor mechanism force in subjects with flexible pronated feet.
Methods: Fifteen asymptomatic female subjects (21.32±1.66 y, 56.30±6.08 kg, 159±6.3 cm) participated in the study. Excessive subtalar pronation was determined by measuring the resting calcaneal stance position (RCSP) in the frontal plane during weight bearing. A neutrally aligned foot was defined as having an RCSP between 2° of inversion and 2° of eversion. On the other hand, a flat foot had an RCSP of more than or equal to 4° of eversion. Both kinetic and kinematic data were collected using a six-camera motion analysis system and a single force plate. Three successful barefoot walking trials were recorded at self-selected speeds. The extensor mechanism force and the adductory component of the free moment (ADD FM) were calculated. The correlation between the ADD FM and the knee extensor mechanism force was examined using the Pearson correlation test.
Results: The Pearson correlation analysis showed a high positive correlation between the ADD FM and the extensor mechanism force ($r=0.917$, $P<0.001$).
Conclusion: Excessive subtalar pronation, along with a possible impairment of the tibial rotation-knee flexion mechanism, may affect the extensor mechanism force at the knee joint.

From a clinical perspective, the possible biomechanical linkage between the knee and the foot complex in the physical examination and treatment of patients should be considered.

Introduction

The subtalar joint (STJ) permits the foot to act both as a flexible mechanism and as a rigid structure during propulsion. This enables the foot to adapt to uneven surfaces as well as facilitate the transmission of forces [1]. Due to its oblique axis, STJ causes a foot pronation (combination of eversion, abduction, and dorsiflexion) directly after initial contact with the ground during normal walking [2,3]. This could be an effective mechanism for

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shock absorption and foot accommodation during the loading response phase of gait [4].

Excessive subtalar pronation or a flat foot is a highly prevalent malalignment of the foot [5,6]. It is a casual mechanism described in relation to injuries of the lower extremities. Evidence indicates that during a closed kinematic chain movement (such as the stance phase of walking), the frontal plane movement of the ankle-foot complex is transmitted to the tibial bone, inducing a transverse (external/internal rotation) movement in the shank [7]. Additionally, the induced internal rotation of the tibia during such movements is shown to be coupled with knee flexion [8]. Any factor affecting the foot's normal function or structure may disturb this coupling mechanism [9]. Excessive foot pronation links the subsequent chain of events and potentially causes an interruption at the knee joint by impairing the normal tibial rotation-knee flexion coupling [8].

Some studies have concluded that excessive pronation could increase the possibility of knee complications and injuries [10-12]. However, other studies failed to show this connection [13,14]. Moreover, several studies have focused on the mechanical consequences of subtalar hyper-pronation on the kinematic chain [15-19]. But less attention has been given to possible kinetic changes due to this malalignment. Abnormal force distribution has been shown as a risk factor for degenerative changes and soft tissue injuries on the knee joint [20]. Since the extensor mechanism is the primary contributor to the knee joint reaction force [21], it seems essential to quantitatively assess the forces produced by this musculature as a valid indicator of force distribution around the knee joint [22].

Theoretically, any alteration in tibial rotation could potentially affect knee movement and, hence, force generation. Therefore, we sought to investigate the correlation between the FM as an indicator of the horizontal plane kinetic variable and the generation of force at the knee joint in functional flat footed subjects. In the present study, the maximum ADD FM was considered. This is the component of the FM that acts to resist toeing out in the initial phase of stance [23]. In our previous study, we found that the peak ADD FM is significantly higher in females with subtalar hyper-pronation in comparison to a control group with normal foot alignments [24].

Considering that the subtalar joint motion is inherently triplanar [25], compensations would be possible in other planes due to its altered motion. Therefore, we believe that the ADD FM in flat-footed subjects will increase along with sagittal plane changes of the kinetic chain (details on how we calculated the ADD FM and the extensor mechanism force are in the Methods section).

The present study hypothesized that excessive subtalar pronation, along with a possible impairment of the tibial rotation-knee flexion mechanism, could affect the knee joint soft tissue structures due to changes in the extensor force mechanism. Providing such a linkage could be clinically relevant during gait analysis procedures. It could also be used as a biomarker for assessment/prediction of knee joint impairment. This novel evidence adds to our understanding of the impact of a functional flat foot on the kinetic chain of the lower extremity from a biomechanical perspective.

Methods

Subjects

Fifteen asymptomatic female subjects (aged 18–30 years) with functional flat feet participated in this study. The subjects were selected after a complete lower extremity clinical examination. Prior to participating, all subjects were informed about the nature of the study. They also signed an informed consent form approved by the Human Ethics Committee of Shiraz University of Medical Sciences. A convenience sampling method was used.

The following were the inclusion criteria for participants: normal range of hip, knee, ankle, and metatarsophalangeal joints motion (based on goniometric assessments); normal (grade five) strength in the major lower extremity muscles (as rated by manual muscle testing conducted by the same examiner); normal hamstring muscle length; being self-ambulatory; and having bilateral flat-foot. Subjects with functional a flatfoot were determined by measuring the resting calcaneal stance position (RCSP) in the frontal plane during weight bearing. This measurement has been shown to have a high intra-rater as well as inter-rater reliability [26]. A neutrally aligned foot was defined as having an RCSP between 2° of inversion and 2° of eversion, while a flatfoot had an RCSP of more than or equal to 4° of eversion [27]. Moreover, a Feiss line test was used to define subjects with a flexible flatfoot [28].

The Feiss line is a straight line from the medial malleolus, through the navicular bone, to the center of the first metatarsal head, assessed during rest and weight-bearing situations [29]. A subject with a pronated foot was labeled flexible if the navicular bone was positioned under the line only in a weight-bearing condition [30,31]. This test has been shown to have high intraday, intra, and inter-tester reliability [32].

The following were the exclusion criteria: functional or structural orthopedic maladies that would prevent a normal stance phase of walking(such as a limb length discrepancy greater than 1 cm); excessive knee hyper-extension; abnormal knee varus or valgus; chronic pain due to structural or functional problems in lower-extremity bones, ligaments, or menisci; neurological ailments affecting the gait (such as neuropathy or other sensory disturbances); or any past history of orthopedic lower limb surgery. All objective measurements of the study were performed by the same experimenter for all subjects. This was done to avoid any possible inter-examiner discrepancies.

The study had a cross-sectional design.

Measures and Procedures

Ground reaction force (GRF) data was collected using a sampling of a single force plate (Kistler Instrument®, Switzerland) at 240 Hz. Kinematic data was collected using a sampling of a six-camera motion analysis system

(Proreflex, Qualisys®Ltd., Sweden) at 120 Hz.

To measure the anthropometric data and subsequently build a six-degrees-of-freedom model, retro-reflective calibration markers, of 19 mm diameter, were placed on the following anatomical points: highest point of the iliac crests; anterior and posterior iliac spines; center of greater trochanters; medial and lateral femoral condyles of the dominant lower extremity; medial and lateral malleoli; first and fifth metatarsal heads; fifth metatarsal base; and the center of the calcaneus (Figure 1).

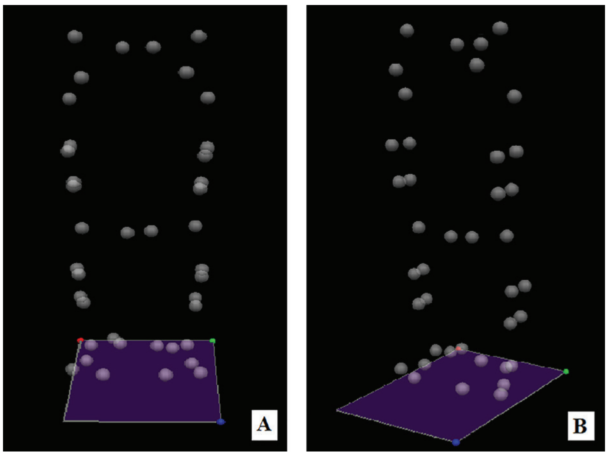

Figure 1: Marker Placement. A: Anterior View; B: ObliqueView

Two sets of cluster markers, containing four tracking markers secured on a polyform material, were placed on the lateral distal of one-third of the shank and on the lateral one-half of the thigh in order to track the movements of desired segments [33].To capture a static trial for the purpose of model building, subjects stood on the force plate and assumed a normal posture for a few seconds.

Following multiple practice trials, three acceptable barefoot walking trials at a self-selected speed were recorded. To promote a natural gait pattern, subjects were instructed to not look down while walking and to maintain visual contact with a fixed point located at the end of the walkway at eye level. Only the trials where the dominant foot of subjects landed on the force plate without disturbing their gait were considered for further analyses. To determine the dominant foot of a subject, they were asked to kick a ball. The foot they used to kick was considered to be the dominant foot [34].

Data was synchronously recorded with QTM software (Qualisys®Ltd., Sweden). All subsequent analyses were performed offline in Visual 3D software (C-Motion®Inc., USA).

Data Analysis

Raw data was filtered using a fourth-order low-pass Butterworth filter with a cut-off frequency of 6 Hz for kinematic data and 15 Hz for kinetic data [22]. The lower extremity was modeled as a rigid, linked-segment system. A standard Newton–Euler method was used to calculate the knee joint angle.

The moment, M_Z, which acts about a vertical axis at the center of force platform, has two components. The first component (FM) is the torque resulting from friction forces between the foot and the ground. Depending on the direction, positive FM (ADD FM) acts to resist toeing out and negative FM (ABD FM) acts to resist toeing in during stance [23] (Figure 2).

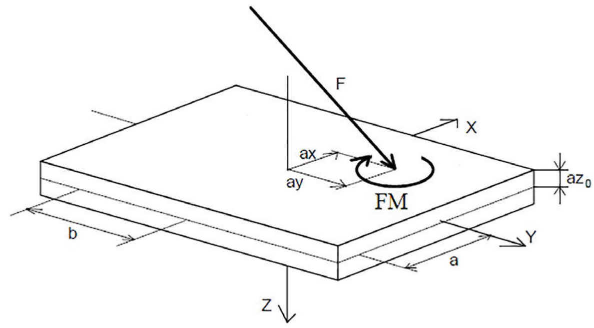

Figure 2: Representation of FM (with kind permission of the publisher: Kistler Instrument AG, Winterthur, Switzerland).

The second component is the moment produced by the resultant shear force acting through the center of pressure (COP). Holden and Cavanagh [35] provided a detailed explanation of the relationship between the two components and MZ. The equation describing the contribution of these two components to the MZ was used to derive the FM from the components of moment and force output from the force platform, according to the manufacturer's instructions (Kistler Instrument AG, Winterthur, Switzerland) [36]. Prior to estimating the FM, all force platform channels were baseline set to a zero value.

$FM=M_Z-(Fy.ax)+(Fx.ay)$ [36]

$Ax=-My'/Fz$ and $ay=Mx'/Fz$

where "M_Z"is the moment about the vertical axis; "ax" is the x-coordinate of the force application point (COP); "Fy" is the ground reaction force in y-direction; "ay" is the y-coordinate of the force application point (COP); "Fx" is the ground reaction force in x-direction;"My" is the plate moment about top plate surface about y- axis;"Fz"is the GRF in z-direction; and "Mx" is the plate moment about the top plate surface about the x-axis[36]. According to the force plate coordinate system, the positive y-axis was in the direction of progression; the positive x-axis was to the left when facing the direction of progression; and the positive z-axis was vertically downward.

The FM was normalized to body weight and height. This was done to reduce the effect of these factors among subjects so that the resultant FM was dimensionless. Peak ADD FM had the maximum positive value of FM during stance.

The extensor mechanism force can be measured from the net internal extensor torque (τ_{em}) on knowing the extensor moment arm [37]:

$$F_{em} = \tau_{em}/d_{em} \quad (1)$$

In this, F_{em} is the extensor mechanism force and d_{em} is the extensor mechanism moment arm.

The following formula was used to calculate d_{em} [22]:

$$d_{em} = 0.0367\, x + 3 \quad (2)$$

In this, x is the knee joint angle in degrees. The knee joint angle was calculated from the difference between the shank segment angle and the thigh segment angle.

On the other hand,

τ_{knee}=*GRF (vertical component)* * d_{GRF} (3)

In this, τ_{knee} is the knee torque and d_{GRF} is the moment arm of the GRF. Knee torque was calculated based on the Visual3D model developed in Visual 3D® software. Assuming that in a semi-static situation, the two torques of extensor mechanism and knee (external flexor torque) are equal ($\tau_{em}=\tau_{knee}$) [22,38], the unknown τ_{em} in equation (1) can be replaced by the knee torque. Therefore, the knee extensor mechanism force was derived from the following formula at the time when the maximum value of ADD FM was reached:

$F_{em}=GRF * d_{GRF}/d_{em}$ (4)

The resultant knee extensor mechanism force was normalized to the body weight.

Statistical Analysis

Data was first analyzed using the Kolmogorov–Smirnov test to recognize a normal distribution. Correlation between the ADD FM and the knee extensor mechanism force was examined using the Pearson correlation. All statistical analyses were performed using SPSS 22.0 (SPSS, Chicago, USA). The level of significance for all tests was set to 0.05.

Results

The demographic characteristics of the participants and the mean values of the studied parameters are included in Tables 1 and 2, respectively.

The Pearson correlation analysis showed a high positive correlation (r=0.917, P<0.001) between the ADD FM (5.94±0.88) and the extensor mechanism force (6.42±0.84) (Figure 3).

Discussion

To our knowledge, few studies have reported the use of the FM in dynamic movement analysis [24,39,40]. In the present study, a positive correlation has been investigated between the ADD FM and the force generation of the knee extensor mechanism in subjects with excessive subtalar pronation. From a biomechanical perspective, the human body has powerful, relative, and synchronous interactions between its segments during walking [41]. Any asynchrony in the motions of the foot and knee segments could potentially result in injury. For example, there could be an injury at the knee joint due to an alteration in the knee joint pattern of movement and force distribution.

Table 1: The demographic characteristics of participants

Variable	Age (year)	Height (cm)	Body Mass (kg)
Mean	21.32±1.66	159±6.3	56.30±6.08
Range	19–25	149–166	45–67

Table 2: Mean values of studied parameters

Variable	Mean	Range
Adductory free moment (*10 -3)	5.94±0.88	3.90–7.90
Extensor mechanism force (N/Kg)	6.42± 0.84	4.57–8.65
Extensor mechanism moment arm (cm)	3.5±0.16	3.25–3.72
External flexor moment (Nm/Kg)	0.088±0.2	0.080–0.32
Knee flexion angle (degree)	14.01 ±4.57	7.14–19.38

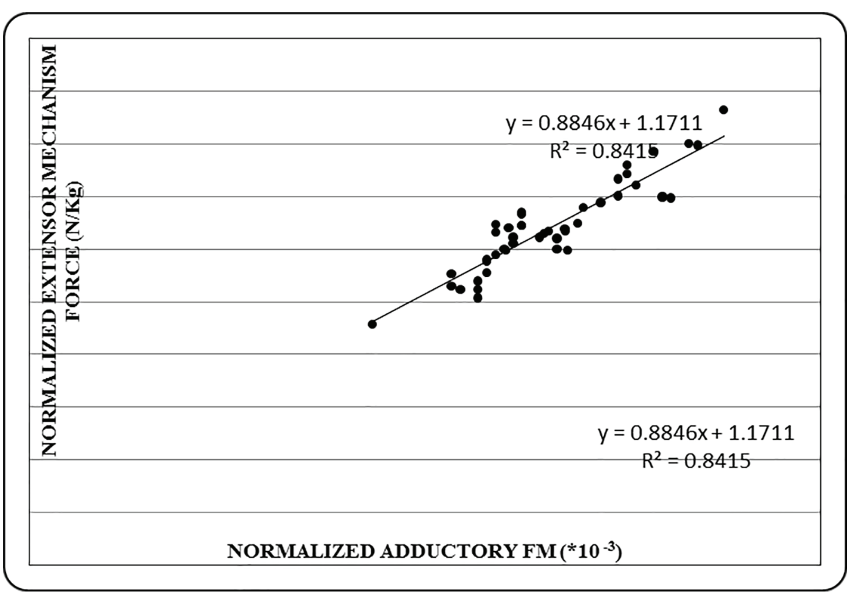

Figure 3: The positive correlation between the normalized ADDFM and the normalized extensor mechanism force

During a normal gait cycle, the knee extension must be associated with tibial external rotation about midstance, in order to preserve congruency of the joint. In a hyper-pronation situation, the femur undergoes excessive internal rotation to maintain relative knee external rotation and compensate for the excessive tibial internal rotation [8].

It has been suggested that in a closed kinematic chain of lower limb movement, the talar adduction in the frontal plane induces knee flexion in the sagittal plane through tibial internal rotation [9]. Therefore, it may be argued that talar adduction as a component of pronation in a close kinematic chain may cause an increase in the ADD FM, along with an increase in the knee flexion angle. Our current finding, therefore, is in agreement with this biomechanical chain of effects. A study by Hunt and Smith showed that subjects with a flatfoot have decreased forefoot adduction at the stance phase [17]. However, our study was not designed to investigate the movements of different foot segments and their connection to the knee extensor mechanism.

Studies have shown that the insertion of an insole, which is believed to compensate for the subtalar hyper-pronation, could reduce the knee joint flexion angle during walking [42-43]. More importantly, it has also been shown that runners with lower arch heights exhibit a greater eversion to the tibial internal rotation ratio in comparison to those with high arches [44]. It could be argued that the horizontal profile of excessive subtalar pronation may contribute to the correlation between the FM in the transverse plane and the knee extensor mechanism force in the sagittal plane.

During normal walking, the extensor mechanism moment arm reaches its maximum amount in 45° of flexion in healthy subjects [45]. It is reasonable to assume that the knee extensor mechanism moment arm in subjects with excessive subtalar pronation may be greater than that among the normal population, because of the increased knee flexion angle in these subjects [8]. This finding is in agreement with our results showing an increase in the extensor mechanism moment arm during the stance phase of walking. Therefore, it can be suggested that the excessive subtalar pronation induces a greater knee flexion angle in the stance phase of gait and, consequently, leads to an increase in the extensor mechanism moment arm magnitude. This finding taken together with ours, can suggest that the observed trend of increase in the knee joint extensor mechanism force is caused by subtalar hyper-pronation.

An increase in the extensor mechanism force may create internal rotational stress at the knee joint, impose a medial rotation at the hip, and cause the knee to face medially (creating a valgus tendency force) [46]. It can also increase the Q angle, pull the patella in a lateral direction, and alter the magnitude of applied force/stress on the soft tissues by changing their insertion alignment on the patella [47]. The possible short-term and long-term clinical significance of such changes are yet to be investigated. On the other hand, the positive correlation between the ADD FM and the extensor mechanism force may introduce the FM as a good predictor tool to determine kinetic changes due to excessive subtalar pronation.

Despite its limitations, the present study was the first to perform a 3D analysis of the influence of transverse plane kinetic changes due to foot morphology on the extensor mechanism force. Anatomical landmark determination for 3D analysis has always been subject to some errors. This was a known limitation of our procedure. Also, all subjects of this study were young females who were not good indicators for flat footed people.

Conclusion

As a result of the rotatory nature of over pronation, the torque applied on the lower extremity can be increased due to excessive subtalar pronation. This can alter the distribution of moment on the entire lower limb. Based on the close kinetic concept, changes of one kinetic variable may lead to changes in other parts of the chain. Considering the ADD FM as a kinetic parameter that controls the horizontal plane movement of the center of gravity, our findings showed that changes in this parameter may lead to sagittal kinetic plane changes of force distribution.

A functional flat foot can alter the force distribution as well as the torque distribution of the entire lower extremity during the stance phase of gait. While this underlying problem affects the motion of the subtalar complex, due to biomechanical linkage with the shank and thigh segments in a closed kinematic chain, an alteration can ensue in the extensor mechanism force. From a clinical standpoint, alterations in the moment arm of a muscle could have devastating consequences in the final torque and, hence, in movement generation. Such a finding could be of clinical relevance since force redistribution on the knee joint and, subsequently, abnormal stresses on soft tissues increase the tendency of musculoskeletal injuries.

Conflict of interest: None declared.

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