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EFFECT OF FEMORAL NECK SHAFT AND ANTEVERSION ANGLES ON HIP CONTACT FORCE

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SUMMARY

Increased femoral neck shaft angle and femoral anteversion are the most common femoral deformities observed in children with cerebral palsy (CP). The hip contact force and muscle forces play crucial roles in the development of these deformities during skeletal growth. The current study aims to find the effects of neck shaft angle and femoral anteversion on hip contact force during gait. In the current work, hip contact force during experimental gait analysis in able-bodied children was calculated using SIMM. Results show that peak hip contact force increases with increasing femoral neck-shaft angle. Depending on the neck-shaft angle, maximum hip contact force was found at 30-50 degrees of femoral anteversion. Future work will be extended to predict the neck shaft angle and femoral anteversion growth tendencies over time resulting from walking in normal children and children with cerebral palsy using finite element analysis.

INTRODUCTION

Children with CP often have normal skeletons at birth [1] but lower limb skeletal deformities often develop thereafter. Coxa valga (increased femoral neck shaft angle) and torsional deformities (femoral anteversion) are frequently observed in children with CP and can lead to hip dislocation [2].

The femoral neck shaft angle (NSA) affects the alignment of the femoral head with the acetabulum and thereby force distribution within the hip joint. The femoral neck shaft angle changes during development; in healthy early infancy the neck-shaft angle is about 150°, in childhood about 140°, in the adult about 125°, and in the elderly about 120° [3]. In children with spastic diplegia, the neck-shaft angle may be increased by 30˚ to 50˚ [4].

Femoral anteversion (FA) is defined as the angle between the neck axis and the condylar axis projected onto the femur's transverse plane [5]. Increased anteversion is therefore a condition in which the femoral neck is anterior to the rest of the femur. Normal femoral anteversion is 40° at birth, 24° at ten years of age, 16° by mid to late adolescent and about 15˚ at skeletal maturity [6–8]. In patients with spastic diplegia the femoral anteversion is commonly 20˚ to 50˚ greater than normal [4].

A previous study shows that at the moment of peak contact force, altered neck shaft angle has only a minor effect on the loading configuration of the hip [9].

The aim of this study is to determine the combined effects of increasing neck shaft and femoral anteversion angles on hip contact force during normal walking.

METHODS

The subjects considered in this study were 10 normal children between 5 and 17 years of age. For this abstract, only results from one 10-year old girl are illustrated and discussed.

Gait analysis was performed using an 8-camera motion analysis system (Vicon MX40) with 2 force platforms (Kistler). From the experimental data, a generic musculoskeletal model (SIMM, Musculographics Inc.) including 88 muscles was scaled based on the marker positions during a measured static pose, to accommodate the anthropometry and weight of the subject. The hip joint was defined as a ball and socket joint. Major hip muscles considered in the analysis were gluteus maximus, medius and minimus divided into anterior, medial and posterior compartments, adductor longus, adductor brevis and adductor magnus, pectineus, iliacus, psoas, quadratis femoris, gemellus, piriformis, tensor fasciae latae, gracilis, semimembranosus, semitendinosus, biceps femoris long head, sartorius and rectus femoris. A deform tool was used at the proximal femur to allow variation of the neck shaft angle and femoral anteversion. The ranges considered in the analysis for neck shaft and femoral anteversion angles are 125°-170° and 10°-60° respectively.

The optimal muscle activation patterns for normal walking were determined by performing inverse dynamic analysis and the static optimization tool in SIMM, which solves the muscle forces that minimize the sum of squared muscle stresses. The calculated optimal muscle activation together with external forces was used as input for a second inverse dynamic analysis to calculate the hip contact force. The peak contact force, generally occurring at contralateral toe-off, as well as computed muscle forces at this instance were analyzed.

RESULTS AND DISCUSSION

The peak hip contact force was extracted from the simulation results and its variation with NSA and FA anteversion in all directions is shown in Figure 1.

With high NSA, the posterior component of peak hip contact force increases is highest at 30-40 degrees of FA. With low NSA, the peak hip contact force is highest at high FA angles.

The inferior component of peak hip contact force, by far the largest, increases with increasing femoral neck-shaft angle. Depending on the neck-shaft angle, maximum hip contact force was found at 30-50 degrees of femoral anteversion.

The lateral component of peak hip contact force decreased slightly with increasing NSA and FA.

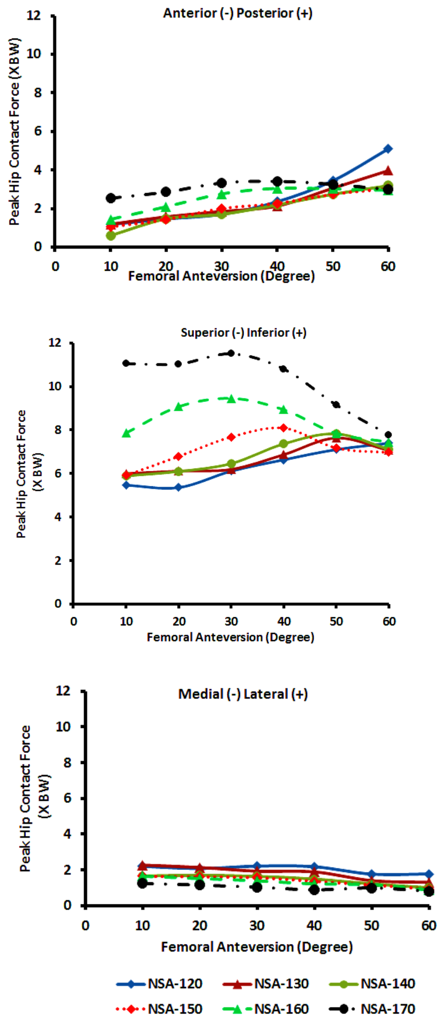

Figure 1: Peak hip contact force on the femur (at contralateral toe-off) in all directions as a function of femoral anteversion, at different femoral neck-shaft angles.

The peak gluteus medius (anterior) muscle force is shown in Fig. 2. This muscle force was highest with large NSA. At lower NSA, anterior gluteus medius muscle force was highest at large FA angles. Many other muscle forces showed a similar trend. The reduced muscle force with increased deformation corresponds with the observed reduction of hip contact force.

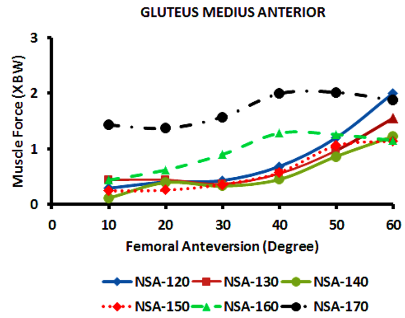

Figure 2: Peak gluteus medius (anterior) muscle force as a function of femoral anteversion, at different femoral neck-shaft angles.

To follow a typical progression of hip deformities in children with CP, at low NSA but mid-to-high FA angles, one can speculate whether the high posteriorly-directed hip contact forces contribute to the frequently-observed increase in FA during growth, and likewise, whether the low laterally-directed contact forces contribute to an increase in NSA. Future studies combining musculoskeletal modeling with finite element analysis on experimental data in this patient population may help to answer these questions.

CONCLUSIONS

This study showed that combined variation in neck shaft angle and femoral anteversion alter the peak hip contact force magnitude and direction. Further studies includes the prediction of neck shaft angle and femoral anteversion growth tendencies in normal children and children with cerebral palsy due to walking loading using finite element analysis.

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