VARIATION IN HIP CONTACT FORCE DURING GAIT DUE TO NECK SHAFT ANGLE AND FEMORAL ANTEVERSION

Priti Yadav1, M.Tech., Eva Pontén2, MD, PhD, and Elena M Gutierrez-Farewik1,2, PhD
1 Department of Mechanics, Royal Institute of Technology, Stockholm, Sweden.
2 Department of Women’s and Children’s Health, Karolinska Institutet, Stockholm, Sweden

1. 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) and femoral anteversion (FA) angles change during development in typically-developing (TD) children, and are commonly 20˚ to 50˚ greater in children with CP [3]. The hip contact force and muscle forces play a crucial role in the development of these deformities during skeletal growth. Our hypothesis is that the forces on the femur during walking can influence growth and deformity development. The aim of this study is to evaluate how varying neck shaft angle and femoral anteversion affect hip contact forces during gait in TD children.

2. Method
The subjects considered in this study were 10 TD children between 5 and 17 years of age. For this abstract, results from only 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.) was scaled to each subject. The hip joint was defined as a ball and socket joint. A deform tool was used at the proximal femur to allow variation of NSA and FA. The ranges considered in the analysis for NSA and FA are 125˚-150˚ and 15˚-60˚ respectively. The expected muscle activation patterns for normal walking were determined by performing inverse dynamic analysis and static optimization in SIMM. The computed muscle activations together with external forces were used as input for a second inverse dynamic analysis to compute the hip contact force. The peak hip contact force as well as computed muscle forces at this instance were analysed.

3. Results
The peak hip contact force on the femur and its variation with NSA and FA are shown in Figure 1. The peak hip contact force was observed at contralateral toe-off position, which is approximately 20% of the gait cycle. The inferior and posterior components of peak hip contact force increase with increasing NSA and FA, while the medial-lateral component remains almost unchanged with increase in NSA and FA. Results also show that peak hip contact force acts posteriorly and laterally for the considered range of NSA and FA.

4. Discussion and Conclusion
Muscles lines and points of action change with varying NSA and FA, thereby changing muscle forces. These changes in muscle forces cause the computed increase in peak hip contact force with increasing NSA and FA. Future studies can be performed to investigate whether this increase in peak hip contact force may inhibit the endochondral growth by reducing the effect of tensile stress on the femoral neck in a growing child.

References

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