**S3 Text. Hip Joint Loading and Gait Energetics in Modern Humans and Early Hominins.**

A general model of pelvic-femoral mechanical loading is shown in S2a Figure [1]. The model is based on an experimental static analysis in one-legged stance [2], and is supported by much in-vivo data and clinical experience. It has been successfully employed in many orthopaedic studies to predict clinical outcomes of altered geometric relationships between abductor muscular and body weight moment arms about the hip, including variation in hip joint reaction forces (JRF) [3, 4]. Predicted magnitudes and orientations of hip JRFs using this model are also consistent with those measured in-vivo using instrumented prostheses [5]. The same principles have been used in finite element models to assess hip arthroplasty failure and femoral neck stress fracture risk [6, 7]. Mediolateral bending of the proximal femoral diaphysis under loading has been confirmed using in-vivo strain gauges, with bending strains increasing with a longer effective femoral neck length [8], again as predicted from the model (S2 Fig).

Both a wider biacetabular breadth and longer femoral neck appear to characterize all early hominins [9], and are consistent with the increased M-L bending strength of the proximal femur also observed in these taxa (S2b,c Fig). According to the model, though, A.L. 288-1 should exhibit more M-L strengthening of this region than early *Homo* (S2c Fig), yet the opposite is found (Fig 4). A reduction in hip JRF in A.L. 288-1, and australopiths generally, would lead to lower predicted increases in M-L bending loads in the proximal femur [10], which is consistent with their morphology in this region. Such a reduction could have been achieved through greater lateral deviation of the body center of mass (COM) over the stance limb [9, 10].

A recent kinematic study questioned the applicability of this traditional model to gait dynamics, showing using inverse dynamics that the body weight moment arm about the hip can vary from that predicted from biacetabular (inter-hip) breadth [11]. However, the effective mechanical advantage (EMA) of the gluteal abductors (abductor moment arm/body weight moment arm) was actually significantly correlated between the dynamic measurement and the traditional measurement in this study (for walking, r = 0.722, p < .001 (n = 25) using femoral neck length and half biacetabular breadth in the traditional model, our calculations from their data). This indicates that the static model shown in S2a Figure still has validity for estimating abductor force, and by implication hip JRF during stance.

Because abductor force constitutes only a minor percentage of total energetic output during normal gait (in straight-line, level walking) [11, 12], changes in the EMA of the abductors, within normal limits, have little effect on locomotor cost [11]. However, based on clinical studies, major alterations in positioning of the body COM over the hip joint, as argued here for A.L. 288-1, would probably have had significant effects on locomotor energetics. Studies of subjects with exaggerated lateral deviation of the body COM during gait due to compromised gluteal abductor function show marked increases in cost of transport. Many of these studies have been carried out on individuals with spinal cord injuries resulting from myelomeningocele (spina bifida), which affects lower limb muscular function to varying degrees. Lesions above the L4 spinal level (thus affecting the gluteal abductors) result in a gait pattern characterized by elevation of the swing side limb accompanied by lateral flexion of the trunk over the stance side, positioning the COM over the stance limb during walking [13-15]. Cost of transport is more than twice that of normal individuals, with a large portion of the increase attributable to these compensatory motions (as deduced from comparisons with costs associated with lower spinal injuries not affecting the gluteal abductors) [15-17]. In line with this, providing external lateral stabilization to patients with incomplete spinal cord injuries significantly reduces cost of transport [18].

None of these specific gait abnormalities would have characterized early hominins, of course. But if A.L. 288-1 (and other australopiths) walked with much greater lateral deviation of the COM than is typical for normal modern humans, this would have entailed a significant metabolic cost. Note that this inference is not based on relative pelvic breadth alone, which has been shown not to affect locomotor cost (at least in modern, relatively narrow-bodied humans) [11], but rather on the total morphological pattern of australopiths, including hip joint size and femoral neck and proximal diaphyseal morphology. While some studies have questioned the association between strain magnitudes and diaphyseal cortical bone remodeling [19], a large number of other experimental and observational studies support this inference [20-23].

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