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Structural behaviour and strain distribution of the long bones of the human lower limbs

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ABSTRACT

Although stiffness and strength of lower limb bones have been investigated in the past, information is not complete. While the femur has been extensively investigated, little information is available about the strain distribution in the tibia, and the fibula has not been tested *in vitro*. This study aimed at improving the understanding of the biomechanics of lower limb bones by: (i) measuring the stiffness and strain distributions of the different low limb bones; (ii) assessing the effect of viscoelasticity in whole bones within a physiological range of strain-rates; (iii) assessing the difference in the behaviour in relation to opposite directions of bending and torsion. The structural stiffness and strain distribution of paired femurs, tibias and fibulas from two donors were measured. Each region investigated of each bone was instrumented with 8–16 triaxial strain gauges (over 600 grids in total). Each bone was subjected to 6–12 different loading configurations. Tests were replicated at two different loading speeds covering the physiological range of strain-rates. Viscoelasticity did not have any pronounced effect on the structural stiffness and strain distribution, in the physiological range of loading rates explored in this study. The stiffness and strain distribution varied greatly between bone segments, but also between directions of loading. Different stiffness and strain distributions were observed when opposite directions of torque or opposite directions of bending (in the same plane) were applied. To our knowledge, this study represents the most extensive collection of whole-bone biomechanical properties of lower limb bones.

1. Introduction

Measuring the stiffness, strength and strain distribution of bones is extremely important to understand bone biomechanics (Fung, 1980), bone formation and adaptation (Lanyon, 1980; Fung, 1990), osteoporosis (NIH, 2000), and fractures (Rockwood et al.,1991).

In some cases a theoretical approach was taken to explain the structural stiffness of the femur, exploiting straight (Toridis, 1969; Cristofolini et al., 1996a) or curved beam theory (Raftopoulos and Qassem, 1987; Fabeck et al., 2002). However, such simplified approach cannot be used to investigate subtle effects. (Martens et al., 1980) measured *in vitro* the torsional stiffness and strength of 46 femurs and 37 tibias (however, specimens were not tested in pairs, and only some of the femurs and tibias came from the same donor). Later, (Martens et al., 1986) tested to failure 15 pairs of femurs in bending. Failure of the proximal femoral metaphysis has often been investigated *in vitro* (e.g.Yang et al., 1996; Lochmüller et al., 2002; Cristofolini et al., 2007). In all such studies, only structural properties were investigated, while the strain distribution was not measured. When the strain distribution was investigated in the femur, in most cases a single loading configuration was used (e.g. Field and Rushton, 1989). A very detailed study on femur strains, although limited to a single specimen, is (Huiskes et al., 1981): they applied different loading configurations to a human femur instrumented with 100 strain gauges. Later, (Cristofolini et al., 2009) measured strains in 12 pairs of human femurs (11 strain gauges in the proximal metaphysis), with 6 different loading configurations. Also, the strain distribution in the tibia has sometimes been measured (Gray et al., 2008).

Viscoelasticity of bone tissue has been demonstrated at the tissue-level (Lakes and Katz, 1979). The Young modulus increases by 10% when the strain-rate is increased by 3 orders of magnitude (Raftopoulos et al., 1993). Most creep takes place in the first seconds, and accounts for typically 5–10% of the strain immediately after load application (Sasaki et al., 1993). However, due to

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the limited viscoelasticity of bone, elastic models are often used to describe cortical (Carter and Spengler, 1978; Fung, 1980) and cancellous bone (Martens et al., 1983; Keaveny et al., 1994). In fact, when cancellous bone was tested at different rates, viscoelasticity became obvious only at very low strain-rates (Guedes et al., 2006). When whole bones are investigated, the practical effects of viscoelasticity are questionable (Cherraf-Schweyer et al., 2007). For instance, Cristofolini et al. (2009) showed that creep over 30 seconds. (most physiological motor tasks take place in a shorter timespan) accounts for only 0.1–3.0% of the initial strain value. In fact, in most Finite Element (FE) models, bone is modelled as a linear material (Helgason et al., 2008). Only recently strain-rate-dependent material properties were implemented in FE models (Helgason et al., 2008).

This review shows some limitations of the current knowledge. Firstly, while the structural behaviour and strain distribution in the femur has been extensively studied (e.g. (Huiskes et al., 1981; Field and Rushton, 1989; Yang et al., 1996; Lochmüller et al., 2002; Cristofolini et al., 2007; Cristofolini et al., 2009)), limited information is available for the tibia: the stiffness was measured in several specimens (Martens et al., 1980; Cristofolini and Viceconti, 2000; Heiner and Brown, 2001), but the strain distribution was measured only proximally (Gray et al., 2008). To our knowledge, no biomechanical properties have been measured for the fibula. Moreover, most of the published studies on lower limb bones focus on single bones, not on entire sets of bones from the same donors. In addition, the practical effect of viscoelasticity on the structural behaviour and strain distribution of whole bones is unclear. It is not ascertained whether different loading rates within the physiological range cause a different response, and if a linear and symmetric mechanical behaviour should be expected in long bones.

The aims of the present study were to:

- Measure the stiffness and strain distributions of the different low limb bones from the same donors;
- assess if there is any significant effect of viscoelasticity on the structural behaviour and strain distribution in whole bones for physiological strain-rates;
- assess if the structure and material properties cause any difference in relation to the direction of the applied load, especially considering opposite directions of bending and torsion.

2. Materials and methods

The structural stiffness and strain distribution of the proximal metaphysis and the diaphysis of the femur, and the diaphysis of the tibia and fibula were investigated.

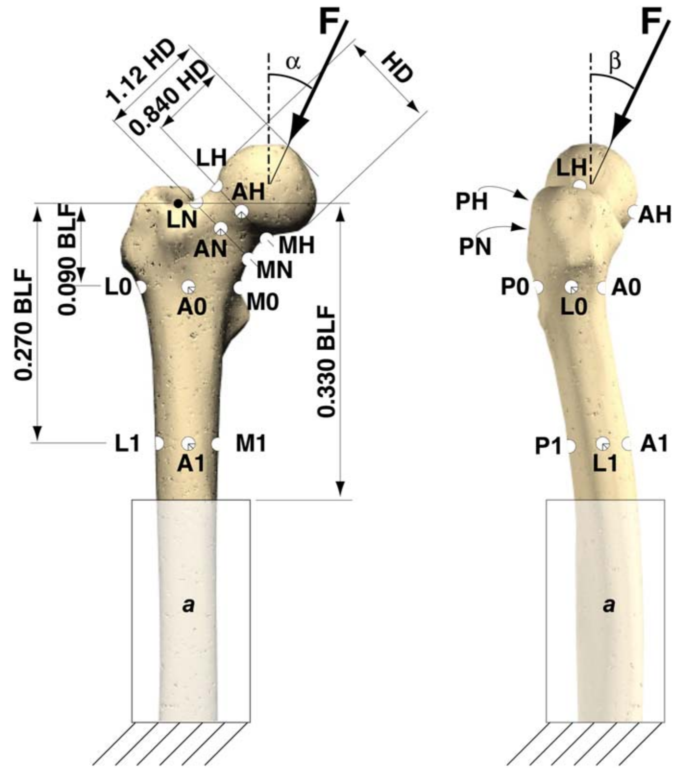

Fig. 1. Anterior and lateral views of the proximal metaphysis of a right femur. The position of the strain gauges is reported: 4 around the head, close to the articular cartilage (AH, LH, PH, MH); 4 around the neck, distal to the previous ones (AN, LN, PN, MN: space on the lateral side was insufficient to host one additional strain gauge); 4 around the proximal diaphysis, just below the lesser trochanter (A0, L0, P0, M0); 4 around the proximal part of diaphysis (A1, L1, P1, M1: they were the same as in the femoral diaphysis, Fig. 2). To enable scaling between specimens, all lengths were defined as a fraction of the head diameter (HD) or of the biomechanical length of the femur (BLF, defined in Fig. 2). The femur was held distally using a pot made of acrylic bone cement (a). The pot could be tilted so that the hip joint resultant force (F) was applied at the prescribed angles in the frontal plane (α), and in the sagittal plane (β), as in Cristofolini et al. (2009).

Table 1
Details of the donors.

	Age at death (years)	Donors' height (cm)	Donors' body weight, BW (kg)	Body mass index, BMI (kg/m^2)	Gender
Donor #1	81	165	63	23.1	Female
Donor #2	78	171	64	21.9	Female

Table 2
Anatomical details of the bone specimens analyzed.

	Femur				Tibia and fibula	
	Head diameter, HD (mm)		Biomechanical length, BLF (mm)		Biomechanical length, BLT (mm)	
	Right	Left	Right	Left	Right	Left
Donor #1	47.5	47.5	427	427	362	364
Donor #2	46.5	47.8	415	412	351	346

The diameter of the head of the femur was measured five times along different directions; the average head diameter, HD, was computed as in (Cristofolini et al., 2009). The 'biomechanical length' of the femur (BLF, see also Fig. 2) was defined as in Cristofolini (1997); the 'biomechanical length' of the tibia and fibula (BLT, see also Fig. 3) was defined as in Conti et al. (2008).