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Measuring Dynamic Stiffnesses of Preloaded Distal Phalanges in Vibration - Test Bench Validation and Parameter Study

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ABSTRACT

An experimental vibration test bench was built for measuring the dynamic stiffnesses and dissipated power densities of preloaded distal phalanges undergoing vibration. The aim of this test bench was to analyse the effects of vibration frequency, static preloading, and vibration excitation amplitude on local biodynamic response. Prior to implementation, the test bench was validated by comparison with a tension–compression testing machine and a reference dynamic mechanical analyser. A measurement study was then conducted on a group of 20 subjects. The mean dynamic stiffness showed that the mechanical behaviour of the index finger distal phalanx is similar to that of a complex amorphous polymer: it exhibits frequency-related stiffening with a rubbery plateau, a glassy transition zone, and a glassy state. The static preloading condition considerably modifies the dynamic response of the phalanx, as well as the dissipated power, which is significantly greater when the preloading is high. An amplitude-related softening phenomenon, similar to the Payne effect for rubber, was also revealed. This can be explained by the thixotropic character of the extracellular matrix of the distal phalanx soft tissues.

Relevance for industry: extensive exposure of the hand–arm system to regular vibration may lead to various disorders and injuries, due in part to changes in mechanical quantities, such as dynamic stress, strain, or dissipated power density, arising from the propagation of such vibration. Nowadays, the direct measurement of this biodynamic response inside soft tissues is still extremely challenging. A way to assess the overall mechanical effects of these local quantities on the human finger is to measure and analyse both the macroscopic stiffness and the dissipated power of fingers.

Keywords: hand–arm vibration, energy dissipation, Raynaud's phenomenon, phalanx dynamic stiffness, soft tissue damping

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1 **1. Introduction**

Sustained exposure to excessive vibration levels may cause a number of pathological
disorders of a vascular, osteo-articular, or musculo-tendinic order (Bovenzi, 1990; Griffin,
1996; Matoba, 1994). These pathologies are generically referred to as vibration syndrome
or hand-arm vibration syndrome (HAVS).

6 Multiple epidemiological studies (Åström et al., 2006; Chetter et al., 1997; Kattel and 7 Fernandez, 1999; Narini et al., 1993) supported by physiological and histological 8 observations of the acute effects of hand-arm vibration (Griffin, 2012; Krajnak et al., 2012) 9 have shown that these disorders are partly related to the frequencies emitted by the 10 machine held by a worker and to other co-factors, such as working posture and pushing and 11 gripping forces.

12 Measurements of the biodynamic responses of the hand-arm system enable the 13 analysis of the influence of such factors, with the aim of better understanding the mechanisms that induce these vibration disorders. For several years, multiple research 14 15 studies have focused on estimating the biodynamic responses of the upper limb; most of 16 these have involved selecting mechanical impedance (Driving Point Mechanical Impedance, 17 DPMI) as the biodynamic response of subjects gripping a vibrating handle (Aldien et al., 18 2006; Burström and Lundstrom, 1998; Gurram et al., 1995; Marcotte et al., 2005). These 19 types of measurement very often provide overall information on the hand-arm system. 20 They are better suited to the study of bone-joint or muscle-tendon pathologies than to that 21 of vascular disorders situated in the finger (digital) blood vessels. Therefore, in order to 22 focus more specifically on vascular pathologies, such as Vibration White Finger or 23 Raynaud's syndrome, it is necessary to collect accurate knowledge of the biodynamic 24 response of finger-transmitted vibration (Griffin, 1996). There have been several attempts 25 to obtain more local experimental data with the help of an instrumented handle by 26 measuring, for example, the distributed impedance in the palm-fingers system (Dong et al., 27 2006, 2005a), or by plotting vibration transmissibility diagrams on the dorsal face of hands 28 (Noel, 2011). These methods are likely to accurately provide overall information on the 29 actual use of hand-held rotary machines. However, it has proven difficult to fully determine 30 the relative contribution of each influencing factor (contact area, push/grip forces, posture, 31 input vibration amplitude, etc.) for a high frequency condition, and some findings even 32 contradicted each other (Burström, 1997; Gurram et al., 1995; Marcotte et al., 2005). 33 However, a number of studies (Adewusi et al., 2008; Dong et al., 2008a) concluded that the 34 biodynamic responses of the hand-arm-finger system measured from instrumented 35 handles are often subject to errors at high frequencies (> 300 Hz). Alternatively, the results 36 are not reproducible between laboratories owing to the measuring bias introduced by the 37 vibrating handle itself at a high frequency (producing problems of mass cancellation at high 38 frequencies, loss of signal coherence, vibration noise related to the handle assembly, etc.). 39 To potentially overcome the previous drawbacks, an alternative might consist of directly 40 measuring the local biodynamic responses at the fingertips by using specific devices. For a 41 number of years, the assessment of the local mechanical impedance of the fingers has been 42 the objective of a number of different research fields, such as of course the pathological 43 effects of finger-transmitted vibration (Lundström, 1984; Mann and Griffin, 1996), but also 44 the field of human tactile sensitivity (Moore and Mundie, 1972; Moore, 1970; Wiertlewski 45 and Hayward, 2012), the design of haptic systems (Chai-Yu and Oliver, 2005; Kern and 46 Werthschützky, 2008), and determining mathematical models of the skin dynamic response 47 (Gulati and Srinivasan, 1997; Hajian and Howe, 1997). The common procedure was to 48 locally vibrate the fingertip with a small-sized probe in contact with the skin for two main 49 phalanx boundary conditions: i) clamped fingernail (Gulati and Srinivasan, 1997; 50 Lundström, 1984; Moore and Mundie, 1972), and ii) free fingernail (Hajian and Howe, 1997; 51 Mann and Griffin, 1996; Wiertlewski and Hayward, 2012). When measuring the local finger 52 biodynamic response, it was easier to carefully control the influencing parameters, such as 53 the probe surface contact and the static-preload, than when assessing the distributed hand-54 finger system with an instrumented handle. The disadvantage of this method is that it does

55 not reproduce the use of hand-held rotary machines accurately. However, at a high-56 frequency condition, it is reasonable to expect the trends of the biodynamic responses to be 57 comparable for whichever test method used, because these trends reflect the mechanical 58 behaviour of the local soft tissues, which may be weakly dependent on those of the entire 59 hand-finger system (Wu et al., 2006). Several studies have been carried out to determine in 60 a nearly comprehensive way the influence of a number of parameters on the local finger 61 biodynamic response, but some of these studies have produced contradictory findings, 62 especially regarding the preloading effect (Lundström, 1984; Moore and Mundie, 1972). 63 Moreover, some researchers pointed out certain potential measurement inconsistencies 64 leading to an overestimation of the inertial effects (Dong et al., 2004b; Wu et al., 2006).

65 The purpose of our study was to describe, validate, and use in parameter analysis form 66 a new experimental vibration test bench for measuring the local biodynamic responses of 67 the distal phalanx. This test bench was designed to provide accurate, reliable, reproducible 68 results at frequencies between 20 Hz and 500 Hz. We aimed at characterizing the 69 mechanical dynamic behaviour of fingertip soft tissues at a high-frequency (> 50 Hz) 70 condition by reducing the effects of other disturbing factors, such as muscle activity and 71 inertial forces produced by the bones or fingernail, as much as possible. Inertial effects were 72 minimized by clamping the fingernail (Wu et al., 2006). To avoid muscle activity as much as 73 possible, the static preloading was performed in a passive way by compressing the fingertip 74 between its support and the probe instead of actively pressing the indenter by using the 75 muscles and tendons of the finger. We first describe the experimental setup, its related 76 metrological equipment, and the signal processing parameters. The deconvolution method 77 for driving the electrodynamic shaker in acceleration is presented. A phase calibration 78 procedure was applied to cancel phase mismatch due to electronic devices. The phase angle 79 accuracy was quantified. Then, the measurement bench was validated by comparison with 80 the results obtained from a tension-compression machine and a reference dynamic 81 mechanical analyser. Finally, a measurement study was performed on 20 subjects. The

82 study parameters were the vibration amplitude and quasi-static preloading. The influence 83 of the vibration frequency on the dynamic stiffness or the mechanical power dissipated by 84 the distal phalanx was analysed before focusing on the effects of the preloading and level of 85 excitation vibration. These results were then examined from a rheological standpoint in an 86 attempt to provide a number of additional physiological explanations.

87 2. Method

88 2.1. Apparatus

89 The experimental setup for measuring the static and dynamic stiffnesses of preloaded90 distal phalanges is shown in Fig. 1.



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Fig. 1. Experimental apparatus: a) laser telemeter, b) control accelerometer, c) force sensor,
d) impedance head, e) shaker, f) aluminium disk, g) safety arm, h) moving support, i) phalanx
support, j) contact indenter, k) hand-arm rest, l) overview of the subject forearm positioning.

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96 The seated subjects place their right forearm on an aluminium platform (Fig. 1 item k). 97 This holder of adjustable height is fitted with four lateral stops with adjustable positions at 98 the elbow and wrist. The purpose of these stops is to ensure appropriate positioning of the 99 subject's forearm and prevent intentional or unintentional movements that could 100 potentially disrupt the measurement. The middle, ring, and little fingers grip a 30 mm 101 diameter plastic cylinder screwed to the holder and adjustable to the subject's morphology. 102 The index finger distal phalanx is adhered (using double sided tape) to the inside face of a 103 curved steel support (Fig. 1 item i). This support is screwed to a piezoelectric force sensor 104 (model B&K 8200) (Fig. 1 item c) rigidly fixed to a steel block. The force sensor is used to 105 estimate the output effort between the fingernail and its support. A piezoelectric 106 accelerometer (model B&K 4705B002) (Fig. 1 item b) with a low measurement range (70 107 m.s⁻² sine wave peak) is positioned near the force sensor to check that the vibration level is 108 sufficiently low so as not to disturb the output force measurement. The distal phalanx is 109 vibrated using an electrodynamic shaker system (model B&K 4809) (Fig. 1 item e) 110 generating a 10 to 20 kHz bandwidth signal and sustaining sine wave peak dynamic forces 111 of a maximum of 45 N. The electrodynamic shaker is fixed to a support micrometer that is 112 controlled manually (Fig. 1 item h). This allows the quasi-static preloading of the distal 113 phalanx by moving the 7 mm diameter cylindrical indenter (Fig. 1 item j). The indenter is 114 screwed to an impedance head (model B&K 8001) rigidly mounted to the vibratory shaker 115 shaft (Fig. 1 item d). The impedance head sensor enables the measurement of both the input 116 force applied by the cylindrical probe to the skin and the input acceleration. The 117 compression distances used to estimate the static stiffnesses from the force/displacement 118 curves were measured using a laser telemeter (model ILD 1700-200 from Micro-Epsilon) 119 with a high resolution of 12 μ m (Fig. 1 item a). The telemeter laser beam impacted a 4 mm 120 thick aluminium disk (Fig. 1 item f) fixed to the shaker moving shaft.

121 The data acquisition, control, and computing operations were performed using 122 MATLAB software with specific modules. A conceptual diagram of the metrological 123 equipment with the different interconnections is shown in Fig. 2.

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Fig. 2. Experimental test bench layout with equipment and interconnections; → source output
 voltage; → measured on NI 9234; → measured on NI 9239.

133 2.2. Biodynamic responses

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134 The dynamic behaviour of the distal phalanges was characterized by the input and 135 output dynamic stiffnesses, $K_{dyn}^{Input}(v)$ and $K_{dyn}^{Output}(v)$, computed conventionally using the 136 cross-spectrum method:

$$K_{dyn}^{Input}(\nu) = -4\pi^2 \nu^2 \frac{S_{F_{I},A}(\nu)}{S_{A,A}(\nu)}$$
(1)

$$K_{dyn}^{Output}(\nu) = -4\pi^2 \nu^2 \frac{S_{F_{O,A}}(\nu)}{S_{A,A}(\nu)}$$
(2)

in which $S_{F_I,A}(v)$ is the cross-power spectral density between the input force $F_I(t)$ and the 137 138 acceleration A(t), both measured with the impedance head; $S_{F_0,A}(v)$ is computed from the 139 output force $F_0(t)$ acquired with the back force sensor and acceleration A(t). $S_{A,A}(v)$ is the 140 power spectral density of acceleration A(t). v was the frequency and t the time. The cross-141 spectra and auto-spectra were calculated for a 2048 Hz sampling frequency using Welch's 142 averaged periodogram method and Hanning windowing, a 50% overlap and a 4 Hz spectral 143 resolution. The recorded time signal duration was 5 s. Time signal coherences were 144 measured to ensure that their values were close to unity in the frequency band used in the 145 study. For the input dynamic stiffness, the mass cancellation method (Håkansson and 146 Carlsson, 1987) was implemented in the spectral domain to overcome added mass effects 147 induced by the indenter and the sensor seismic mass. The stiffnesses were chosen to 148 characterize the biodynamic response instead of the more conventional mechanical 149 impedance, because it is possible to explain the findings in the framework of experimental 150 rheology, in which the stiffness is often used to characterize materials. This is also the 151 reason for the output stiffness to be preferentially selected in the analysis of the findings. 152 The fingernail was clamped so that the test bench would simulate classical rheological 153 experimental devices as closely as possible. We therefore adopted the vocabulary used in 154 rheology. The real part of the dynamic stiffnesses was termed the storage modulus and the 155 imaginary part the loss modulus.

In addition to the input and output dynamic stiffnesses, the mechanical power exchanged between the cylindrical indenter and distal phalanx could be estimated. If we assume that the force exerted by the indenter on the phalanx is exerted at a point, the mean power \mathcal{P}_{ext} generated by this force and applied to the distal phalanx can be conventionally expressed by frequency-based integration of the cross-spectral power density $S_{F_I,V}(\nu)$ between the force F_I and velocity V at the point of application:

$$\mathcal{P}_{ext} = \int_{-\infty}^{+\infty} S_{F_I,V}(\nu) \, d\nu \tag{3}$$

162 The expression shown in Eq. (3) can easily be rewritten with the input dynamic 163 stiffness $K_{dyn}^{Input}(v)$ and acceleration autospectrum $S_{A,A}(v)$:

$$\mathcal{P}_{ext} = \int_{-\infty}^{+\infty} \left[\frac{1}{(2\pi\nu)^3} Im \left(K_{dyn}^{Input}(\nu) \right) \right] S_{A,A}(\nu) \, d\nu \tag{4}$$

164 In the special case of pure harmonic excitation with a frequency v_0 and an amplitude 165 a_0 , the mean power generated by the external forces becomes:

$$\mathcal{P}_{ext} = \left[\frac{1}{(2\pi\nu_0)^3} Im\left(K_{dyn}^{Input}(\nu_0)\right)\right] \frac{{a_0}^2}{2}$$
(5)

166 Since the kinetic energy variation is zero over a vibration cycle, the mean power 167 generated by the external forces is, in absolute terms, equal to the mean power of the 168 internal forces. Eq. (5) thus represents the mean power dissipated by the phalanx during a 169 vibration cycle. This dissipated power is, of course, controlled not only by the excitation 170 amplitude, but also by the imaginary component of the input dynamic stiffness, or similarly 171 by the real component of the input DPMI, since these two quantities are deduced from each other by division by a factor of $2\pi i \nu$, $i = \sqrt{-1}$. The influence of different parameters 172 173 (frequency, amplitude, preloading) on energy dissipation was investigated in previous 174 studies by the real component of the DPMI (or equivalently by the imaginary component of 175 the stiffness) either for the local (Moore and Mundie, 1972; Wiertlewski and Hayward, 176 2012) or the global (Besa et al., 2007; Burström and Lundstrom, 1998) biodynamic 177 responses of the fingers-hand-arm system. However, we introduced the so-called 178 *acceleration power kernel* written $\mathcal{Ker}(v)$ and expressed as:

$$\mathcal{K}er(\nu) = \frac{1}{(2\pi\nu)^3} Im\left(K_{dyn}^{Input}(\nu)\right)$$
(6)

179 for characterizing the properties of energy dissipation. The acceleration power kernel was 180 chosen by analogy with other studies (Dong et al., 2008b, 2006), where it was used to 181 propose a vibration-power-absorption-based weighting alternative to the existing ISO-5349 182 filter (ISO 5349-1:2001, 2001). Indeed, Eqs. 4 and 5 clearly show that power absorption is 183 proportionally linked to the square of acceleration through the acceleration power kernel. 184 As for the real component of the DPMI or the imaginary component of the stiffness, the 185 acceleration power kernel enables the characterization of the power dissipation 186 independently of the acceleration spectrum applied to the fingertip. However, it explains 187 the dissipative properties of the fingertip assuming the same level of input acceleration. 188 Based on previous studies (Dong et al., 2005c, 2004a), where a new ISO-5349 filter was 189 proposed from the vibration power absorption density in the whole finger, we could use the 190 acceleration power kernel measured locally at the fingertip as a new candidate for ISO-5349 191 weighting under high frequency.

193 2.3. Generation of required target input acceleration

The shaker and its power amplifier (Fig. 2) were assumed to behave as a linear system characterized by impulse response h(t) between voltage V(t) and acceleration A(t). The aim was to establish a method of computing the voltage V(t) that would produce the required target acceleration A(t) at the system output. To solve this problem, we implemented a filtering method based on a restricted least squares regression, also termed the regularization method (Norcross, 2009). The Tikhonov regularization method was applied to compute the voltage $\tilde{V}(t)$, which is expressed as:

$$\tilde{V}(t) = \arg\min_{V(t)} \{ \|A(t) - h(t) * V(t)\|^2 + \lambda \|c(t) * h(t)\|^2 \}$$
(7)

in which arg min is the argument of the minimum, $\|.\|$ is the Euclidean standard, λ is a positive scalar parameter, and c(t) is a function to be defined.

Fig. 3 illustrates an example of implementation of this methodology and shows the measured acceleration generated by the shaker input voltage computed from the desired target acceleration. The *rms* acceleration measured in this way was 10.3 m.s⁻², which closely approximated the expected value (10 m.s⁻²).



Fig. 3. Power spectral density of input acceleration transmitted by the cylinder indenter to the
 phalanx; ----- 10 m.s⁻² rms target acceleration; — measured acceleration (10.3 m.s⁻² rms)
 generated by shaker input voltage computed from desired target acceleration.

211 2.4. Phase angle measurement accuracy

212 To minimize the measurement errors of both the complex dynamic stiffness and 213 mechanical dissipated power, it is necessary to reduce the phase angle mismatch generated 214 by the acquisition system by as much as possible. These phase angle errors are linked to the 215 sensors (negligible in this case owing to the high resonance frequency of the sensors), the 216 channel-to-channel phase mismatch of the input acquisition module, and the phase shift 217 caused by the charge amplifiers (intrinsic electronic, digital, or analogue filtering, etc.). The 218 phase calibration was a two-step procedure (Fig. 4) consisting of: i) measuring the channel-219 to-channel phase mismatch, and ii) evaluating the phase shift of the charge amplifier by 220 using a standard capacitor to simulate an ideal charge input.









Fig. 4. Sketches showing procedure used for system phase recalibration: (a) channel-to-channel
phase angle mismatch measurement; (b) phase angle mismatch of charge amplifier measurement.

224 To verify the phase angle adjustment procedure, the output stiffness phase angle $K_{dvn}^{Output}(v)$ was measured for a steel spring. Given the very low damping of this type of steel 225 226 spring, its dynamic stiffness phase angle should be virtually zero. To ensure the correct 227 positioning of the spring, the phalanx support was replaced by a flat support and the 228 cylindrical indenter by a disk-shaped indenter. The graph of the dynamic stiffness phase angle $K_{dyn}^{Output}(v)$ obtained for our INRS (Institut national de recherche et de sécurité) test 229 230 bench spring compared to that of a theoretical spring is shown in Fig. 5. As shown in the 231 figure, it remains confined within a \pm 2° band. Other measurements were taken for different 232 springs, which provided similar results (Noel, 2015).



Fig. 5. Phase angle of dynamic output stiffness for control spring; ——INRS test bench;
theoretical spring.

236 2.5. Test bench validation

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237 To validate the measurements of the static and dynamic stiffnesses, the INRS test bench 238 results were compared with those provided by reference experimental systems: a quasi-239 static tension-compression machine (Wolpert Testatron 20 TTZ) and a viscoanalyzer 240 (Metravib VA 2000). The tested sample comprised a 15 mm diameter, 14 cm high 241 polyurethane foam cylinder with mechanical characteristics approximating those of the 242 index finger phalanx. Tests were conducted using the tension-compression machine at a 243 displacement speed of 1 mm.min⁻¹ (quasi-static load can be considered). The force-244 displacement curves subsequently obtained are shown in Fig. 6. The two force-245 displacement curves exhibited similar shapes. The INRS test bench results were higher than 246 those obtained from the reference tension-compression machine from a compression of 1 247 mm upwards. The relative error between these two measurements was 9.2%, 6.5%, and 248 8.5% at compressions of 2 mm, 4 mm, and 6 mm, respectively. The difference between the 249 two devices was linked to the fact that, for the tension-compression machine, the 250 compression speed was very low (it can be considered to exhibit quasi-static behaviour), 251 while for the INRS test bench, the compression speed ranged between 20 and 40 mm.min⁻¹ 252 (caused by the time taken by the experimenter to compress the finger). Therefore, because

of the relaxation phenomena associated with the viscoelasticity properties of the finger and the higher compression speed of the INRS test bench than that of the tension-compression machine, the stress measured did not reach its quasi-static low limit. This could explain why the force-displacement curves measured with the two different devices were not identical.



Fig. 6. Quasistatic test bench validation: force versus displacement of reference polyurethane foam
 sample; ——INRS test bench; ----- reference tension-compression machine.

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261 The dynamic stiffness of the polyurethane foam was then measured using a commercial 262 viscoanalyzer. The material had previously been preloaded at a 15% compression ratio 263 (ratio of compressed length to initial length), then vibrated by a sine wavefront generated in 264 10 Hz increments between 20 and 100 Hz. This viscoanalyzer was coupled with a 0.5 mm 265 constant displacement at 9 frequencies. Measurements were taken under strictly the same 266 experimental configurations as those of the INRS test bench. The increase in dynamic 267 stiffness measured with these two metrological systems is shown in Fig. 7 (a). As shown in 268 the figure, the increase of the dynamic stiffness with respect to the frequency is consistent 269 for the two test benches. The stiffness plateau at 50 Hz, 60 Hz, and 70 Hz is visible on both 270 curves. The relative difference between the measurements was of the order of 5%. The 271 phase angle curves of these two sets of dynamic stiffnesses are shown in Fig. 7 (b). As 272 shown with the dynamic stiffness, the general shapes of the two curves were similar and the

273 relative error between the two phase angles remained below 6%. In conclusion, the INRS
274 test bench is a reliable and accurate experimental setup enabling the measurement of the
275 dynamic stiffnesses of preloaded phalanges in terms of both magnitude and phase angle.



Fig. 7. Dynamic test bench validation: (a) dynamic stiffness magnitude of reference polyurethane
foam sample; (b) phase angle of reference polyurethane foam sample; — with INRS test bench;
with commercial dynamic mechanical tester.

279 **3. Subjects and experimental set-up**

- 280 In accordance with the French law, the study protocol was reviewed and approved by
- the French state organism allowing biomedical research on human subjects.
- 282 3.1. Description of subject cohort

Static and dynamic stiffnesses were measured on 20 healthy, non-smoking, subjects (8 women and 12 men) aged between 19 and 39 years, and with a normal skin sensitivity threshold for their age (monofilament size 3.61 in Semmes-Weinstein test). Their distal phalanx thicknesses were measured using the laser telemeter (Fig. 1 item a). Their distal phalanx widths and index finger lengths were measured from three-dimensional images obtained by scanning each subject's hand with a laser scanner. Anthropometric markers common to all subjects and easily identifiable on these images were established in order to standardize the dimensional measurement. The distal phalanx width was defined as the width in the frontal plane (parallel to the palm of the hand) of the distal interphalangeal joint, and the index finger length was defined as the distance between the end of the finger and the metacarpophalangeal joint. The forearm volume was evaluated by simply measuring the quantity of water displaced when the subject placed his/her forearm up to the elbow in a water-filled container. Table 1 summarizes the different average anthropometric characteristics recorded for the subjects.

297 Table 1

298 Overall and forefinger specific anthropometric characteristics for cohort subjects; 299 μ = mean and σ = standard deviation; BMI = Body Mass Index.

	Overall	charact	eristics			Forefinger characteristics			
	Age [year]	Size [m]	Weight [kg]	ВМІ [-]	Forearm volume [l]	Distal Phalanx Thickness [mm]	Distal Phalanx Width [mm]	Index Finger Length [mm]	
μ	22.9	1.7	66.3	22.4	1.2	13.5	17.2	78.1	
σ	5.5	0.1	8.4	2.6	0.3	1.5	1.5	5.5	

300 3.2. Experimental procedure

The quasi-static stiffnesses were computed as the derivatives of the output forces F_0 with respect to the cylindrical indenter displacements measured with the laser telemeter. The dynamic stiffnesses (Eq. (1) and Eq. (2)) were estimated for the following two experimental parameters: the *rms* values of the cylindrical indenter input acceleration and the quasi-static preloading. The input accelerations were generated as described in Section 2.3. Their spectral shape is shown in Fig. 3 and their *rms* values are listed in Table 2. 307 The static preloading state can generally be defined in a variety of ways, such as: i) the 308 cylindrical probe iso-displacement, ii) the iso-force applied by the probe on the fingertip 309 (this is the definition most often selected in other studies (Dong et al., 2005b; Lundström, 310 1984; Lundström and Burström, 1989)), and iii) the iso-compression rate relative to the 311 fingertip thickness. The methods that would involve specifying the preloading in relation to 312 absolute compressions or forces were not selected, because a similar compression or force 313 would produce, in two different subjects, two distinct forces or compressions, whose 314 difference was not necessarily due to variations in mechanical properties, but to initial 315 geometrical differences. For example, an 18 mm thick phalanx compressed by 3 mm would 316 not be in the same static preloading state as a similarly compressed 11 mm thick phalanx. 317 Moreover, a level of preloading expressed as a degree of compression with respect to the 318 total phalanx thickness again would not define a reference state of preloading common to 319 all subjects (for a given fingertip thickness, the flesh and bone thicknesses could be different 320 for different subjects). Therefore, as the reference state of preloading we adopted the 321 operating points (x_i, F_i) on the force F_i – displacement x_i curve, which provided the same 322 static stiffness for all subjects *i*. The static stiffness is the macroscopic resultant of the load 323 and deformation fields. Hence, we considered it a relevant indicator for defining a reference 324 state of preloading. We could therefore compare the dynamic stiffnesses measured for each 325 subject for an equivalent static load in an iso-stiffness sense.

Table 2 summarizes the 20 parameters (5 accelerations × 4 states of preloading) used
for the dynamic stiffness measurements.

328 Table 2

329 Summary of setup parameters used for the dynamic stiffness measurements.

Indenter shape and dimensions	Acc	Acceleration level					Quasi-static stiffness			
	[m.s ⁻² <i>rms</i>]				[N.mm ⁻¹]					
Cylinder with diameter of 7 mm	2	5	10	15	20	2	5	8	10	

330 **4. Results**

331 4.1. Static stiffness

332 We plotted static stiffnesses as a function of compression for five subjects in Fig. 8 (the two extremes and three other representatives—the curves of the other subjects are similar 333 334 but are not shown for a better visualization). The static stiffness curves for all subjects have 335 the same general shape that can be divided into two main parts. At low compression, the 336 stiffness is small with a low slope. Beyond a compression of 2.5-3.5 mm, the stiffness 337 increases quickly with higher compression values. This behaviour is similar to that 338 observed in certain elastomers or some polyurethane cellular foams (Riande et al., 2000). 339 The results of the present study are in good agreement with those reported in other works 340 (Serina et al., 1997; Wu et al., 2003) in which high dispersion existed over the subjects and 341 the stiffness had the same order of magnitude.



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Fig. 8. Example of static stiffness for five subjects; ---- subject with high stiffness from low
compression; -- subject with high stiffness from high compression

345 4.2. General frequency behaviour and inter-subject variability

The mean storage and loss moduli and the output stiffness $K_{dyn}^{Output}(v)$ averaging phase are shown in Fig. 9 for the following experimental set-up: a vibration excitation amplitude of 5 m.s⁻² rms and a quasi-static preloading equivalent to a static stiffness of 8 N.mm⁻¹. These
 parameters were selected because the obtained dynamic stiffness allows characterization of



the dynamic behaviour of the distal phalanx very distinctly.

Frequency [Hz] Fig. 9. Three basic ways to present the dynamic output stiffness for the experimental condition: acceleration 5 m.s⁻² *rms* and preload 8 N.mm⁻¹; the left axis is assigned to the stiffness modulus, the right axis to the phase angle; — storage modulus; — loss modulus; — phase angle; — linear interpolation of the storage modulus with an offset of -5 N.mm⁻¹ for better visualization.

358 Overall, the stiffness storage modulus increases continuously with frequency. The first 359 phenomenon revealed is a stiffening of the distal phalanx as the vibration excitation 360 frequency increases. The storage modulus increases from 16 N.mm⁻¹ at 25 Hz to 37 N.mm⁻¹ 361 at 500 Hz. The distal phalanx is twice as stiff at 500 Hz as at 25 Hz. The real part of the 362 stiffness exhibits three distinct frequency zones. The first zone exists between 25 and 125 363 Hz where the stiffness is approximately 20 N.mm⁻¹ and has moderate variation (the gradient 364 is 2 N.mm⁻¹ per octave). The second zone lies between 125 and 315 Hz where the storage 365 modulus increases quickly from 21 to 33 N.mm⁻¹ with an average gradient of 9 N.mm⁻¹ per 366 octave. The third zone lies between 315 and 550 Hz (the spectral limit of our

367 measurements), where the stiffness increases rapidly (the gradient is 16 N.mm⁻¹ per 368 octave). The shapes of the stiffness curves in Fig. 9 are similar to those plotted using 369 viscoanalyzer measurements for amorphous polymers (Menard, 2008 p. 26 and Lakes, 2009 370 p. 228). By analogy with rheology, the phalanx stiffness can be considered to be composed 371 of a rubbery plateau of low stiffness up to 125 Hz, followed by a region of steep rise in 372 stiffness between 125 and 315 Hz, called the glassy transition, and finally, a glassy state of 373 higher stiffness (more than twice that of the rubbery plateau). The dynamic stiffness phase 374 angle has two local maxima at 150 and 450 Hz that are associated with two points of 375 inflection in the real part of the stiffness. In the field of rheology, the first local maximum is 376 called the α peak or relaxation α and is associated with the point of inflection in the glassy 377 transition zone. The second local maximum is called the β peak or relaxation β and is 378 related to the point of inflection in the stiffness rise in the glassy state.

Input and output stiffnesses exhibit a rubbery plateau followed by a glassy transition zone before entering an ultimate glassy state (Fig. 10 (a) and (b)). Input and output storage moduli are very close up to a frequency of approximately 125 Hz (Fig. 10 (a)). Input and output rubbery plateaus are then quite similar. Above 125 Hz, the input storage modulus is less than the output modulus. Unlike the phase of the output stiffness, that of the input stiffness increases continuously up to 400 Hz where the relaxation β is distinct.

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394 Fig. 10. Comparison of input and output dynamic stiffness for the following experimental conditions: 395 acceleration 5 m.s⁻² rms and preload 8 N.mm⁻¹; (a) storage modulus; (b) phase angle; —— storage 396 modulus of input stiffness; —— storage modulus of output stiffness.

397

398 We quantified the inter-individual variation using the 95% confidence interval to gain 399 an understanding of the output stiffness measurement dispersion. The mean output 400 stiffness and 95% confidence interval are shown in Fig. 11 for the following experimental 401 conditions: an acceleration level of 5 m.s⁻² rms and a quasi-static preloading of 8 N.mm⁻¹. In 402 this case, the half-width of the 95% confidence interval is 20%, and it has the same order of 403 magnitude under all other experimental conditions.





407 4.3. Acceleration power kernel

408 The acceleration power kernel is shown in Fig. 12 for the following conditions: 409 acceleration level = $5 \text{ m.s}^{-2} \text{ rms}$ and quasi-static preloading = 8 N.mm^{-1} .



Frequency [Hz]
Fig. 12. Acceleration power kernel for experimental condition: acceleration 5 m.s⁻² rms and
preloading 8 N.mm⁻¹; — mean value; 95% confidence interval.

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414 The acceleration power kernel decreases quasi-linearly with respect to frequency. The 415 interpolated straight line slopes at approximately -7.6 dB/octave. In other words, the 416 acceleration power kernel (and hence, the dissipated power for harmonic excitation) is 417 divided by six, when the frequency doubles. As described in Sections 4.4 and 4.5 of this 418 paper, this behaviour is exhibited regardless of the excitation amplitude and static 419 preloading. We can therefore establish the following empirical rule: when a phalanx is 420 preloaded and harmonically vibrated by a cylindrical indenter (diameter: 7 mm) under iso-421 acceleration, doubling of the frequency leads to a six-fold reduction in the power dissipated 422 by the distal phalanx.

When the vibration excitation is sinusoidal, the power dissipated by the distal phalanx is related to both the acceleration amplitude and frequency, as shown in Eq. (5). Based on this relation, iso-power dissipation charts can be plotted with respect to the *rms* sinusoidal

426 acceleration level and its frequency. Examples of these charts are provided in Fig. 13. The 427 acceleration power kernel used for plotting these charts corresponded to that of the 428 following experimental condition: acceleration level = $5 \text{ m.s}^{-2} \text{ rms}$ and quasi-static 429 preloading = 8 N.mm⁻¹. In the following sections, we show that this acceleration power 430 kernel depends on the quasi-static preloading but depends weakly on excitation amplitude 431 at frequencies below 80 Hz. Hence, these abacuses are suitable only above this frequency. 432 Charts of the same type but different from those shown in Fig. 13 can be plotted for each 433 static preloading.



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Fig. 13. Charts of dissipated iso-power under pure harmonic vibration for a preloading of 8 N·mm⁻¹.
The dissipated power was computed from the acceleration power kernel based on the following
experimental condition: acceleration 5 m.s⁻² *rms* and preloading 8 N.mm⁻¹.

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These charts are especially of interest because they provide an overall view of the dissipated power variation with respect to the acceleration level and frequency. They also enable quick selection of an acceleration level or frequency combination that yields the given dissipated power. For example, it is easy to find which acceleration level or frequency combination is associated with a dissipated power of less than 0.1 mW. Under the assumption that we can link vibration pathology development to a dissipated power threshold that must not be exceeded, such charts will enable us to select the threshold amplitude or frequency combinations for vibration-induced white finger pathology at frequencies higher than 80 Hz.

4.4. Effects of preloading for a fixed vibration level

(a), (b), (c), and (d) for an acceleration level of 5 m.s⁻² *rms*.

The impact of the quasi-static preloading at a fixed acceleration level can be analysed with regard to the following four quantities: storage modulus, loss modulus, acceleration power kernel, and output stiffness phase angle. These four quantities are shown in Fig. 14



Fig. 14. Effects of preloading for a 5 m.s⁻² rms vibration level; (a) output storage modulus; (b) output
loss modulus; (c) acceleration power kernel; (d) phase angle of the output stiffness;
** 2 N.mm⁻¹; ** 5 N.mm⁻¹; ** 8 N.mm⁻¹; ** 10 N.mm⁻¹.

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The effect of the static preloading on the storage and loss moduli is clearly shown in Fig. 14 (a) and (b). These moduli increase with the static preloading. The preloading influences for the rubbery plateau and the glassy transition zone are not identical. On the rubbery plateau, in addition to the overall increase in stiffness, when the preloading increases, the slope of the interpolated straight line on this plateau increases more 481 moderately that that of the interpolated straight line in the glassy transition zone (an

482 example of these straight lines is provided in Table 3 with the values of the gradients).

- 483 **Table 3**
- 484 Slope of the storage modulus on the rubbery plateau and in the glassy transition zone for various
- 485 preloadings.

Quasi-static stiffness	2	5	8	10	
[N.mm ⁻¹]					
Slope	Rubbery plateau	1	1.5	2	2.5
[N.mm ⁻¹ per octave]	Glassy transition	2	5	9	12

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487 The consequence of an increase in preloading is, therefore, an overall rise in dynamic 488 stiffness with more pronounced effects in the glassy transition frequency zone (the ratio of 489 the slopes for the maximum and minimum preloadings is 6) than on the rubbery plateau 490 (the corresponding ratio is 2.5). This synergy between frequency and static preloading was 491 also confirmed by the results of a variance analysis test in which the frequency and 492 preloading combined term had a p-value much less than 0.05. In addition to the increased 493 stiffness, the loss modulus increases with increase in the preloading; in this case too, the 494 effect is more pronounced in the glassy transition zone. As showed in Fig. 14 (c), the 495 acceleration power kernel is small when the preloading is low. The decrease in the 496 frequency of the acceleration power kernel remains virtually identical regardless of the 497 level of quasi-static pre-load. There is no synergetic effect between the level of preloading 498 and frequency. The acceleration power kernel curves are simply translated when the 499 preloading varies (except at approximately 400 Hz, where the curves differ slightly). We 500 quantified the relationship between the static preloading variation and the power 501 dissipation variation under iso-acceleration. On average, over the frequency band, doubling 502 the static preloading causes twice as much energy dissipation (this ratio varies between 1.8 503 and 2.3 over the spectral bandwidth). Moreover, the output stiffness phase angles shown in

Fig. 14 (d) exhibit two paradoxical effects of the static preloading on frequency mismatch in relaxations α and β : the lower the preloading, the lower are the frequencies of α and the higher are the frequencies of β . In other words, decreasing the static preloading effectively reduces the spectral bandwidth of the rubbery plateau and increases the bandwidth of the glassy transition zone, even when these variations are relatively low, especially in the case of β .

510 To gain a statistical understanding of the results, we plotted the mean storage and loss 511 moduli and the corresponding lower and upper bounds of the 95% confidence interval in 512 Fig. 15 (a) and (b). For both storage and loss moduli, the 95% confidence intervals are well 513 separated for low static preload (2 N.mm⁻¹ and 5 N.mm⁻¹). In contrast, for higher static 514 compressions (8 N.mm⁻¹ and 10 N.mm⁻¹), the 95% confidence intervals overlap. We also 515 developed a general model for the analysis of variance (ANOVA) to determine the 516 significance of the static preloads on the storage and loss moduli (differences were 517 considered significant at the p < 0.05 level). Statistical analysis with the data from all 518 subjects at a fixed vibration amplitude (here, 5 m.s⁻² rms) shows that both frequency (p < r519 (0.0001) and static preload (p < 0.0001) are statistically significant factors for dynamic 520 stiffness. Furthermore, there exists a statistically significant interaction between frequency 521 and static preload (p < 0.03). In fact, as shown in Fig. 15 (a) and (b), beyond 125 Hz, the 522 slope of the dynamic stiffness in the leathery state depends on both the compression level 523 and frequency. In addition, a Tukey's honest significant difference (HSD) multiple 524 comparison test was performed to determine the means of the storage and loss modulus 525 that are significantly different from each other. We found that the means of storage and loss 526 modulus for each compression rate are statistically significantly different from each of the 527 other three. The statistical analyses for the acceleration power kernel and phase angle are 528 not presented here because these two quantities are determined from storage and loss 529 moduli, and hence, the same conclusions as those for the latter quantities can be drawn. To

(a)



Fig. 15. Variability of subjects for a 5 m·s⁻² rms vibration level; (a) output storage modulus;
(b) output loss modulus; <u>2 N.mm⁻¹</u>; <u>5 N.mm⁻¹</u>; <u>8 N.mm⁻¹</u>; <u>10 N.mm⁻¹</u>;
...<u>A</u>. upper bound of the 95% confidence interval; ...<u>V</u>·lower bound of the 95% confidence interval.

535 4.5. Effects of the vibration level for a fixed preloading

The output stiffness loss moduli are shown in Fig. 16 for the four preloadings and fiveacceleration levels listed in Table 2.



543 For the two smallest preloadings (2 N.mm⁻¹ and 5 N.mm⁻¹), only the first four 544 acceleration amplitudes are plotted in Fig. 16 (a) and (b). This because when the excitation 545 amplitude is 20 m.s⁻² rms, the coherence between acceleration and force is too low, and the 546 transfer function cannot be calculated accurately under such conditions. Fig. 16 (a) to (d) 547 clearly show a decrease in the distal phalanx stiffness when the excitation amplitude

548 increases. An analysis of variance highlighted that both frequency (p < 0.0001) and 549 vibration amplitude (p < 0.0003) are statistically significant factors for the storage modulus. 550 There is no synergic interaction between frequency and amplitude (p > 1). Indeed for each 551 vibration amplitude, the storage moduli are simply translated (like an offset effect) without 552 any modification of the shape of the associated curves. A Tukey's HSD test was performed to 553 identify the amplitudes for which the means of the storage modulus are significantly 554 different. It shows significant differences between the means of the storage modulus at high 555 acceleration amplitudes and low levels. The results are listed in Table 4.

556 **Table 4**

Statistical multi-comparisons of the significant differences in the mean storage modulus for several
vibration amplitudes; the grey cells indicate no significant differences, and the white cells indicate
significant differences.

	2 m.s ⁻² <i>rms</i>	5 m.s ⁻² <i>rms</i>	10 m.s ⁻² rms	15 m.s ⁻² <i>rms</i>	20 m.s ⁻² rms
2 m.s ⁻² <i>rms</i>					
5 m.s ⁻² <i>rms</i>					
10 m.s ⁻² rms					
15 m.s ⁻² rms					
20 m.s ⁻² rms					

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561 Fig. 17 (a), (b), (c), and (d) (both linear and logarithmic scales were used) show that the 562 variation in vibration excitation amplitude has an insignificant influence on the acceleration 563 power kernel for all frequencies for 2 N.mm⁻¹ and 5 N.mm⁻¹ preloadings and above 80 Hz for 564 8 N.mm⁻¹ and 10 N.mm⁻¹ preloadings. The acceleration excitation amplitude affects distal 565 phalanx softening or stiffening, but has limited influence on the dissipated vibration power 566 kernel at high frequencies (>80 Hz). In other words, at high frequencies (>80 Hz), the input 567 vibration amplitude does not intrinsically affect the dissipation of the mechanical power of 568 the phalanx regardless of the compression. At low frequencies, the input vibration 569 amplitude has a small effect only for large preloadings.



576 **5. Discussions**

577 5.1. Main potential measurement errors

578 When measuring the output force, the fingertip is placed on a steel shell support. The 579 weight of this phalanx support (11 g) was added to that of the top seismic mass of the 580 piezoelectric element, thereby reducing the resonance frequency of the sensor. However, by 581 using the sensor characteristics in its data sheet and by assuming a mechanical second-582 order system behaviour, we theoretically found that the modified resonance frequency (>10 583 kHz) remained much higher than the maximum frequency of the spectral band considered 584 in our study. Hence, the weight added by the phalanx support caused little disturbance in 585 the output force measurements. Furthermore, the impedance of the foundation on which 586 the force sensor is fixed can lead to discrepancies between the true and measured forces if it 587 is not considered in the force assessment (Braender, 1972). However, in the case of a very 588 heavy foundation (i.e. very low foundation velocity), the measurement error becomes 589 insignificant except near the resonant frequency of the sensor. For each measurement, we 590 conducted a check with the control accelerometer (Fig. 1 item b) placed near the force 591 sensor to ensure that the foundation velocity remains negligible.

592 With regard to the static stiffness assessment, both the fingertip compression i.e. the 593 cylindrical probe displacement and the output static force should be accurately estimated. 594 However, phalanx compression may be inaccurate if the just-in-contact condition between 595 the probe and phalanx is improperly evaluated. Indeed, the probe displacement should be 596 measured with reference to the position at which the indenter and phalanx barely touch in 597 order to prevent initial deformation of the phalanx, which can cause a measurement error. 598 Given the very low stiffness of soft tissues with respect to minor compressions, the just-in-599 contact condition is extremely difficult to estimate and reproduce by the experimenter. 600 Therefore, a simple electronic system that operates at very low currents (<2 mA, which is safe for humans) was developed. When the indenter makes contact with the distal phalanx,an electroluminescent diode lights up.

603 Assessing the quasi-static effort with a piezoelectric sensor is a great challenge. 604 Actually, the measuring system only provided an estimate of the DC component of the force 605 response because of both the metrological limitations caused by the combination of the 606 charge amplifier and piezoelectric sensor and the physical stress relaxation phenomena. 607 The stresses measured dynamically at a given moment were different from those obtained 608 under quasi-static conditions (Wu et al., 2003). Forces were measured at a low loading 609 speed to minimize viscous effects. Conversely, if the measuring time increases, the 610 electronic discharge process (low frequency limitation) and charge amplifier output voltage 611 drift (zero drift) strongly disrupt the force measurement (Gatti and Ferrari, 1999). The final 612 measuring time reflects a balance between the discharge time constant (100 s), charge 613 amplifier output voltage drift (no effect if the measurement time is less than 20 s), and 614 relaxation effect (weak if the measuring time exceeds 10 s based on phalanx tests). Based on 615 a large number of measurements synthesised in the technical report (Noel, 2015), we 616 concluded that if the measuring time is between 10 s to 15 s, the relative error between true 617 static force and the quasi-static force measured with the piezoelectric force output sensor is 618 less than 10%. This is in good agreement with the results of the validation tests as indicated 619 by comparison with a reference compression test machine (Section 2.5). The measuring 620 time selected was, therefore, between 10 s and 15 s.

Moreover, the aluminium disc with the telemeter for displacement measurement, the screw for fixing the impedance head, the cylindrical probe, and the fingertip support were designed such that their first resonance frequencies were much higher than the upper limit of the spectral range analysed in this study. This ensured that no noise was transmitted by these elements to the fingertip because the measured coherence functions between the different mechanical quantities were almost unity.

627

628 5.2. Rheological standpoint and physiological analogy

629 The dynamic stiffnesses of the distal phalanxes exhibit similar behaviour as those of the 630 measured stiffnesses of certain elastomers (Fig 9). This indicates that the macromolecular 631 structure of the conjunctive tissues of the distal phalanx is likely similar to that of 632 amorphous polymers. More specifically, the viscoelastic properties of distal phalanxes are 633 mainly induced by the extracellular matrix of the conjunctive tissues (Vicente, 2012). These 634 tissues occupy the space between the cells, nourish, and support the cells. These tissues are 635 composed of a complex network of macromolecules that can be grouped into three main 636 types (Callen and Perasso, 2005 p. 432-443): i) very voluminous fibrous proteins, i.e. 637 collagen and elastin; ii) less voluminous glycoproteins that ensure the adhesion of the 638 different matrix constituents with the cells or among themselves; and iii) complex 639 carbohydrates (polysaccharides) and glycosaminoglycans that are often attached to a 640 protein to form proteoglycans. They form a hydrated gel that fills the water-retaining 641 matrix. The adhesive glycoproteins and polysaccharides form a gel substance called the 642 ground substance. The most important polysaccharide in the ground substance of the distal 643 phalanx is hyaluronic acid (Feng and Lu, 2011; Vicente, 2012). The hyaluronic acid chains 644 have secondary proteoglycan chains (the versican in the dermis: a special type of 645 proteoglycan) that are linked by adhesion glycoproteins to the constituents of the 646 extracellular matrix, such as collagen and elastin. Because of this complex assembly of 647 macromolecules, the distal phalanxes exhibit a viscoelastic behaviour similar to that of 648 elastomers. The vibration energy dissipation properties are therefore explained (Purslow et 649 al., 1998; Shen et al., 2011) either by molecular rearrangements of different components of 650 the extracellular matrix (more specifically in the collagen and ground substance) or by 651 movements that cause viscous shear forces between the fluid (ground substance) and solid 652 (collagen and elastin) phases.

The softening phenomenon related to an increase in the excitation amplitude (Fig. 16)is known in certain amorphous polymers, especially those with large carbon black content.

655 The phenomenon is called the Payne effect or the Fletcher–Gent effect. This dependence of 656 distal phalanx stiffness on the vibration excitation amplitude can be considered in the 657 context of the thixotropic nature of the ground substance (Feng and Lu, 2011 p.14). The 658 ground substance effectively has the consistency of a gel with the two defining properties of 659 thixotropy (Barnes, 1997): i) the gel is rheofluidifying (shear rate thinning), i.e. its viscosity 660 decreases with increasing mechanical stress. In other words, this gel behaves more as a 661 solid at low stresses and more as a fluid at higher stresses. The macroscopic consequence of 662 this characteristic may be a reduction in stiffness under increased amplitude (relaxation 663 associated with a more fluid nature of the ground substance at high stresses). ii) the effects 664 of this fluidification are not out of phase with stress application, but occur later. Similar to 665 viscoelasticity, this behaviour reflects the dissipative nature of the ground substance.

666 5.3. Physical interpretation of results and comparison with other works

667 A number of works (Reynolds and Angevine, 1977; Sörensson and Lundström, 1992; 668 Wu et al., 2006) have highlighted that low-frequency vibrations can propagate from the 669 fingers to the whole hand-arm system, while high-frequency vibrations (>100 Hz) are more 670 localized in the finger contact area because of the mechanical damping abilities of soft 671 tissues (Wu et al., 2006). Hence, findings may be compared in terms of the trend of high-672 frequency vibrations, regardless of the test methods even if some differences are present. 673 Thus, our results may be considered either from the perspective of local finger biodynamic 674 responses or the responses distributed in the hand-finger system.

The biodynamic response of the fingertip is expected to be controlled by the tissue elasticity up to a frequency of approximately 125 Hz (almost constant stiffness—segment [AB] Fig. 9—with a phase less than 15°). The same behaviour was observed in a number of studies where the frequency of slope failure (point B Fig. 9) varied between 100 Hz and 200 Hz. In Lundström's study (Lundström, 1984), this frequency is approximately 200 Hz, while it is approximately 100 Hz for Kern and Werthschützky (Kern and Werthschützky, 2008), 681 Mann and Griffin (Mann and Griffin, 1996), and Wiertlewski and Hayward (Wiertlewski and 682 Hayward, 2012) and 130 Hz for Moore and Mundie (Moore and Mundie, 1972; Moore, 683 1970). The latter two studies highlighted that the failure frequency at 130 Hz occurs not 684 only at the fingertip, but also in the thenar eminence, the volar forearm, and the thigh. This 685 observation led to the hypothesis that this failure frequency is not specific to the finger but 686 is related to the mechanical behaviour of the human skin. Dong and co-workers (Dong et al., 687 2004c) reported a stiffening of the fingers biodynamic response as measured with an 688 instrument handle at approximately 80 Hz. The leathery transition (segment [BC] Fig. 9) 689 and the glassy state (segments [CD,EF] Fig. 9) were not emphasized by Lundström 690 (Lundström, 1984), but Wu and co-workers (Wu et al., 2006) found some inconsistency in 691 Lundström's measurements at high frequencies. However, the leathery and glassy states 692 were thoroughly measured by Moore and Mundie (Moore and Mundie, 1972) who found the 693 glassy state at approximately 400 Hz, which is in good agreement with our findings. Dong 694 and co-workers (Dong et al., 2004c) also reported changes in the dynamic response at 300 695 Hz and 500 Hz. On a separate note, Liang and Boppart (Liang and Boppart, 2010) measured 696 the Young's modulus of the human skin in the palm and forearm by using dynamic optical 697 coherence elastography. They concluded that the Young's modulus of the skin is almost 698 constant between 100 Hz and 200 Hz, and then increases up to 300 Hz and remains 699 constant up to 400 Hz, beyond which, it rises again. This material behaviour is very similar 700 to those of the storage modulus measured in this study.

Output and input stiffnesses differ from 125 Hz (Fig. 10). Indeed, the input stiffness is mainly controlled by the soft tissues of the fingertip, whereas the output stiffness is influenced by stiffer anatomical elements such as the bone and the nail (because of measurement localisation). Similarly, the input stiffness phase angle is greater than the output stiffness phase angle because the bones and the nails have less damping ability than the soft tissues.

707 The effect of the static preloading on biodynamic responses was studied by Lundström 708 (Lundström, 1984) who found a slight increase in the dynamic stiffness with doubling of the 709 pre-stress, but the preloading level considered in the previous study was much less than 710 that in the present study. In the study of Moore and Mundie (Moore and Mundie, 1972), the 711 preloading was only significant for frequencies less than 130 Hz, whereas in the high-712 frequency region, the effect of static loads was measured. However, Dong and co-workers 713 (Dong et al., 2004c) and Kern and Werthschützky (Kern and Werthschützky, 2008) found 714 that the gripping forces or static loads have considerable effects, but they do not cause 715 dissimilar behaviour between the rubbery plateau and the leathery or glassy state, as we 716 found in the present work. However, the boundary conditions of the fingers in their works 717 were different from those in our study. Their works incorporated inertial effects largely, 718 and this may be the cause of the observed differences.

719 Lundström (Lundström, 1984) studied the effects of input acceleration amplitude on 720 biodynamic responses. He found that the input acceleration amplitude exerts an influence 721 only in low-frequency (<100 Hz) region; however, for the same reason as explained earlier, 722 no definitive conclusions can be draw at high frequencies. In the study of Gurram and co-723 workers (Gurram et al., 1995), the input acceleration amplitudes do not modify the hand-724 arm-system biodynamic response regardless of the frequency range. On the other hand, 725 Marcotte and co-workers (Marcotte et al., 2005) found that the level of input acceleration 726 could change the mechanical impedance at low frequencies only (<100 Hz). In the present 727 study, we showed that the input acceleration amplitude may have a major influence on the 728 storage modulus at all frequencies. We further showed that it affects the dissipated 729 mechanical power slightly only at low frequencies (<50 Hz).

730 6. Conclusions

In this study, an experimental apparatus was built for measuring static and dynamic
stiffnesses in preloaded distal phalanxes. The experimental set-up and test bench were

733 validated by comparing their results with those obtained using a tension-compression 734 machine and a viscoanalyzer. Twenty subjects participated in the test, to provide the 735 statistical data representative of an average index finger. Analysis of the measured dynamic 736 stiffnesses highlighted that the phalanx exhibited mechanical behaviour similar to that of 737 certain amorphous polymers. The phalanx stiffnesses in fact increase with the vibration 738 frequency, exhibiting a rubbery plateau, a glassy transition zone, and a glassy state. Under 739 iso-excitation amplitude and iso-static preloading, the power dissipated in the phalanx is 740 divided by six when the frequency is doubled. The influence of the static preloading and 741 vibration excitation amplitude was subsequently investigated. The higher the preloading 742 state, the higher is the dynamic stiffness with a very pronounced frequency and preloading 743 synergy effect below 125 Hz. The dissipated power is also affected by the level of preloading 744 because doubling this level leads to twice as much vibration dissipation. Additionally, an 745 increase in vibration excitation amplitude causes a decrease in the phalanx stiffness, similar 746 to the Payne effect or the Fletcher-Gent effect encountered in a number of rubbers. The 747 thixotropic nature of the extracellular matrix may be the cause of this decrease in phalanx 748 stiffness. Our measurements also revealed that the acceleration power kernel was not 749 influenced by the vibration excitation level at high frequencies (clearly, the dissipated 750 power is directly linked to the vibration amplitude). Using these kernels, the iso-dissipated 751 power charts were plotted for each static preloading and were used for selecting the 752 frequency and amplitude combinations to realize a target dissipated power. These 753 experimental results will be used for adjusting and validating finite element models of 754 preloaded, vibrated phalanxes.

755

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