HUMAN NECK CHARACTERIZATION UNDER THORACIC VIBRATION – INTER-INDIVIDUAL AND GENDER INFLUENCE

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ABSTRACT

The objective of the present study is to improve knowledge of the dynamic human neck behavior *in vivo*, an essential aspect in rear end impact protection optimization. Neck characterization is performed in the frequency domain by a vibration loading of the volunteer thorax. Inter-individual differences and gender influence are also investigated. The mean modal characteristics of the human head-neck system extracted from the 30 experiments and the related standard deviation are defined by two natural frequencies located at f_1=1.68±0.69 Hz and f_2=7.91±0.74 Hz. For all subjects the first vibration mode was a flexion-extension motion and the deformed mode shape related to the second natural frequency was a head translation also called retraction motion. When male and female were distinguished, only small differences appeared, i.e. for the male, mean value and standard deviation of the two natural frequencies of f_1=1.83±0.96 Hz and f_2=8±0.8 Hz against f_1=1.53±0.28 Hz and f_2=7.83±0.68 Hz for the female group. Muscle effect was also analyzed and it was possible to demonstrate that the contacted condition shows a major increasing of the first natural frequency value, especially for males. The new data recorded on human being *in vivo* constitute new validation parameters for mathematical neck models and dummy necks.

Keywords: Vibrations, Modal analysis, Head-neck system, Rear end impact.

INTRODUCTION

Despite advances in safety devices, neck injuries in traffic accidents, especially non-severe rear impact accidents, continue to be a serious and costly social problem. The high cost of whiplash injury has been extensively documented in several countries Szabo *et al.*, 2002, Szabo and Welcher, 1996. The development of safety measures designed to decrease the incidence of whiplash injuries must be guided by meaningful and reliable human body surrogates. Most injury prevention strategies are based on impact analysis using anthropomorphic crash test dummies or mathematical models. Without proper evaluation of these experimental or computational models against the mechanical responses of the human body, it will not be possible to improve the current state-of-the-art neck injury prevention techniques. Unfortunately the cervical spine is one of the most complex structures in the human skeletal system and its behavior during impact is still poorly understood.

At present there are no less than three crash test dummies dedicated for use in experimental rear impact analysis; the Hybrid III dummy developed by Foster *et al.*, 1977, the BioRID II designed at Chalmers University (Davidsson, 1999) and the RID dummy proposed by TNO in the Netherlands (Cappon *et al.*, 2001). A number of validation studies have been conducted on these dummies against volunteers and against post mortem subject neck responses (Cappon*, et al.*, 2001, Davidsson *et al.*, 1999, Davidsson, 1999, Prasad *et al.*, 1997, Seemann *et al.*, 1986, Siegmund *et al.*, 2001). These studies have demonstrated the limited biofidelity of this human body surrogate under low speed rear impact. Optimization studies of the car-seat-head rest system were also described by Eichberger *et al.*, 1996, Ishikawa *et al.*, 2000, Svensson *et al.*, 1993, Szabo*, et al.*, 2002 and concluded that the safest protective system against whiplash depends on the dummy used, which illustrates the importance of accurate human body surrogate biofidelity.

Typically, numerical or physical neck model validation is conducted against volunteers or postmortem human subjects (PMHS) by comparing the evolution of recorded mechanical parameters over time with the human response corridor. This methodology is limited as it is very difficult to characterize a multiple degrees of freedom system under impact in the temporal domain. These difficulties are well illustrated by the various prototypes of test dummies and the large number of evaluation and comparative studies found in the literature (Cappon*, et al.*, 2001, Philippens *et al.*, 2002, Prasad*, et al.*, 1997). The number of prototype versions and contradictions between study conclusions illustrate how difficult it is to explain some phenomena that are masked within the time

domain. An other illustration can be found in Philippens, *et al.*, 2002 study where "realistic" dummy head kinematics can be observed, but T1 accelerations were out of corridors. The reason for this is that the dummy response has to remain within ranges or corridors with wide tolerance. The evaluation process in the temporal domain is not sufficiently accurate to prescript initial ramps, local peaks and oscillations of the head response that can be of great importance. Despite this critical issue, recent researches in spine biomechanics have improved our knowledge of this complex structure. The limitations listed above illustrate the need for further experimental and theoretical analysis. The purpose of this paper is to apply modal analysis techniques to characterize the human head-neck system *in vivo*. Indeed modal analysis in engineering is non-destructive and used for identification of dynamic structures. In biomechanics the method has been used extensively for bone healing processing and for dynamic characterization of the human head (Hodgson *et al.*, 1967, Stalnaker and Fagel, 1971, Willinger and Cesari, 1990).

In previous studies undertaken at the University of Strasbourg, the experimental modal analysis of the human head-neck system *in vivo* performed by applying an antero-posterior impulse to the head, provided us with a first set of natural frequencies and mode shapes (at respectively 1.3±0.3 Hz and 8±0.8 Hz), which constitute original validation parameters for dummy necks (Willinger *et al.*, 2002). A detailed description of the applied methodology can be found in Meyer *et al.*, 2004 and Willinger *et al.*, 2003, demonstrating how this experiment provided the biomechanical background for dummy and numerical neck model evaluation. It is however questionable if head-neck characterization by frontal head impact is sufficiently representative of neck loading conditions occurring under rear impact, when the head is loaded by inertia. Thorax vibration will also enable us to introduce more energy in the system at higher frequencies.

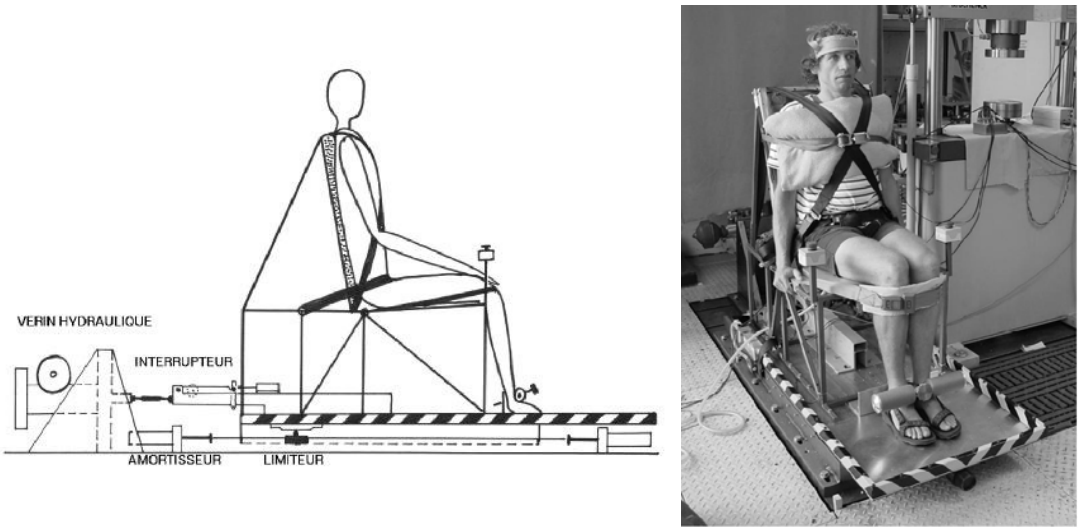
In the present study thorax vibration was used as loading condition for a new experimental neck analysis under relaxed and contracted neck muscle. Main objective therefore is to compare head-neck modal characteristics under head and thorax loading. A second objective is to increase the number of cases and to conduce an inter-individual analysis, especially concerning gender effect.

METHODOLOGY DESCRIPTION THROUGH A FIRST CASE

A total of 30 volunteer subjects were evaluated: 15 women and 15 men. The volunteers are in the range 23-60 years range of age for men, and 25-40 years range of age for women. The new experimental device consists of a carriage on which a rigid seat is fixed as illustrated in Figure 1. The carriage is vibrated by a combination of sinusoidal signals of 5 s of duration in the 0.2-16 Hz frequency range, transmitted by a hydraulic actuating cylinder. The vibration is repeated ten times in order to minimize noise associated with the digitizing effects. The total duration of the test is about 50 second. During the thorax vibration, the head and neck rotations do not exceed a few degrees (≤ 10° for each segment). The recorded input signals are the acceleration of the first thoracic vertebra (T1), and the output signals are the accelerations of the head at the center of gravity level (CG) as well as the acceleration of the Occipital Condyl (OC). The volunteer is belted on the seat in order to prevent a relative movement between T1 and the seat. The detailed components of the device are as follows:

- The vibration exciter is a hydraulic machine (Schenck). The hydraulic cylinder used (PL 10) can reach a static force of 10 kN and a dynamic force of 8 kN for a maximum stroke of 100 mm.
- The carriage consists of a plate guided by two linear bearings. The hydraulic cylinder is related to it by a safety system, which dissociates itself from the carriage when the loading is over 4 g.

Acceleration signals are digitized with a data acquisition using a low pass filter of 100 Hz and signal conditioning machine 12 bit card. After this process all data analysis use the original digital data without any other filtering. Signal acquisition and data processing are obtained with a developed computer program. Two transfer functions are then calculated in terms of transmittance between T1 and CG on one hand, and between T1 and OC on the other hand.



(a) **(b)**

Figure 1 : (a) Representation of the experimental device; (b) A volunteer in position on the experimental device.

The first step however involves confirming the linearity of the head-neck system by calculating the coherence function between the input acceleration signal at T1 (x(t)) and the output acceleration at OC and CG (y(t)). The equation of the coherence function can be written as Equation 1 (Bendat and Piersol, 1971):

Equation 1 $$\hat{\gamma}_{xy}^{2}(w) = \frac{\left|G_{YX}(w)\right|^{2}}{G_{YY}(w).G_{XX}(w)}$$

where G_{XX}, G_{YY} and G_{XY} are the autospectrums and interspectrum of the signals obtained in the frequency domain (Bendat and Piersol, 1971).

The system is generally accepted to be linear if the coherence function remains between 0.8 and 1 in the frequency range analysed. The response signal contains not only the response due to the measured excitation, but also the response due to ambient random excitation. This typical measurement can therefore be characterized as having noise in the measured output signal. Using the principle of least squares to minimize the effect of noise at the output the best Frequency Response Function (FRF) estimator between input at T1 and outputs at OC and CG is given by Equation 2 (Bendat and Piersol, 1971):

Equation 2 $$\hat{H}(w) = \frac{G_{XY}(w)}{G_{XX}(w)} = \left|\hat{H}(w)\right| e^{-j\hat{\Phi}_1(w)}$$

In order to minimize noise associated with the digitizing effects and to improve the transfer function estimator, the time measurement is of 50 s corresponding to 5 seconds per period for 10 periods, which are a combination of sinusoidal functions.

The tests were carried out on healthy volunteers without disease of the cervical spine. Before each test, the volunteers were questioned by filling in a form and informed about experience. Neck muscles influence is conduced with two configurations that are in relaxed and fully contracted situation. In the relaxed condition, the volunteer closes his eyes without more contraction of neck muscles than he needs to maintain the head. In contracted condition, the volunteer keeps his eyes open staring at a point in front of him, and contracts his neck muscles to thwart the movements due to inertial effects of the head.

The next steps of the methodology involving data processing and result analysis, are descript by referring to a first volunteer subject. Figure 2 and Figure 3 report the transfer function and coherence for a given subject respectively under relaxed and contracted condition. The transfer functions between both output accelerations (at CG and at OC) and the input acceleration at T1 are expressed in terms of transmittance using Equation 2. The frequency diagram in Figure 2 and Figure 3 are the Real and Imaginary parts of the transmittance and the related coherence function. The coherence between 0.8 and 1 at frequencies contained in the 2-14 Hz range confirms the linear behavior of the head-neck

system under study conditions, which gives the validation domain of the transfer function.

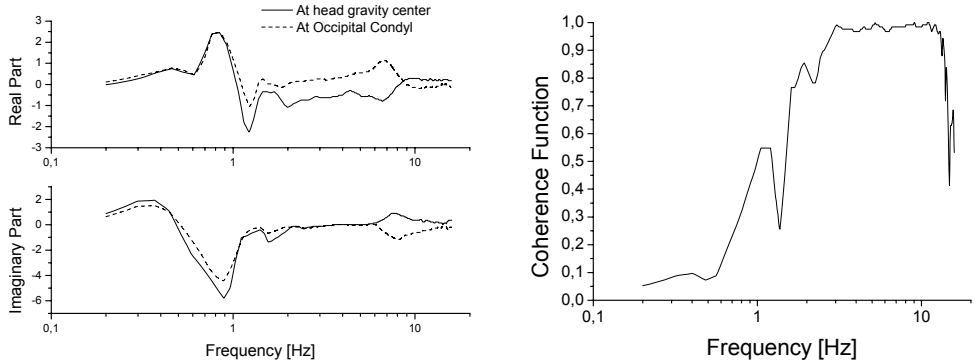


Figure 2 : Representation of the transfer function in terms of real part, imaginary part as well as its coherence function for a subject in relaxed condition.

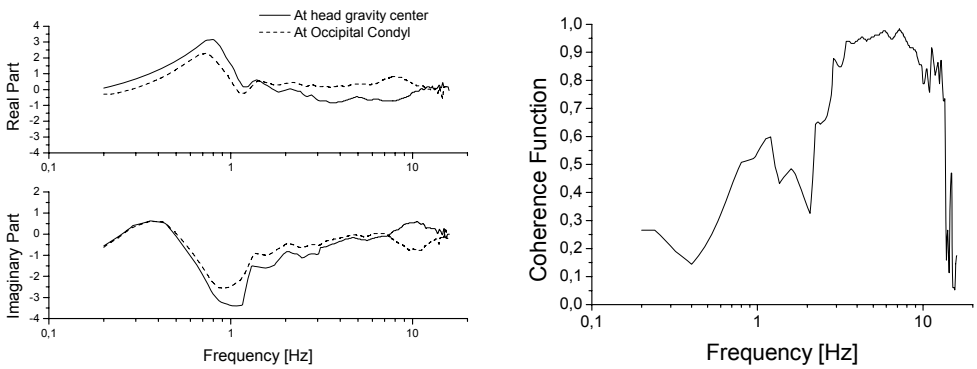


Figure 3 : Representation of the transfer function in terms of real part, imaginary part as well as its coherence function for a subject in contracted condition.

The accurate natural frequencies of the system is calculated using the real part of the transfer functions (Ewins, 1984). This curve reveals at sign change, slopes that indicate a resonance. The frequency is calculated using Equation 3 (Bendat and Piersol, 1971):

Equation 3 $$f_r = \sqrt{\frac{f_a^2 + f_b^2}{2}}$$

Where f_1 and f_2 are the slope limits at each natural frequency on the real part.

Figure 4 illustrates the real parts calculated at head gravity center with the necessary frequency limits for the determination of the natural frequencies for each mode, for the relaxed case.

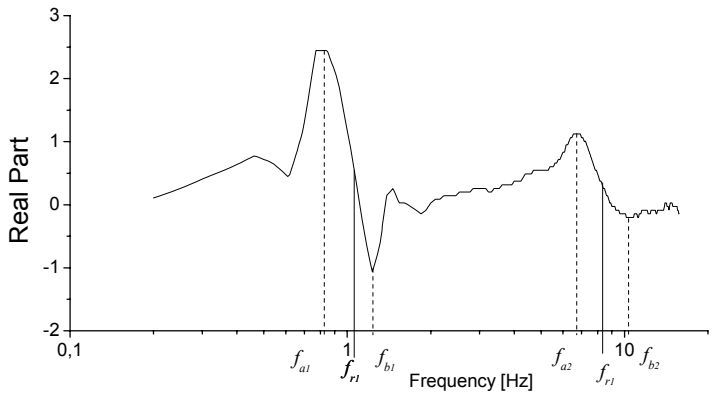


Figure 4 : Determination of the natural frequencies extracted from the real part of the transfer function at CG level for the relaxed condition.

For our first volunteer, a first natural frequency is found at 1 Hz (relaxed) against 1 Hz (contracted), and a second resonance appears at 7.7 Hz (relaxed) against 10.2 Hz (contracted). The real part also enables us to calculate the damping ratio which is defined by Equation 4 Bendat and Piersol, 1971:

Equation 4 $$\eta = \frac{f_b - f_a}{2 f_r}$$

For the first natural frequency, we were not able to extract the damping of the first mode with this methodology due to a lack of energy at low frequencies. Therefore this dampingde was extracted from the head impulse test performed in a previous study, for which energy was concentrated at low frequency. The damping ratio obtained is of 0.3 under both relaxed and contracted situation. In contrast, the damping ratio of the second mode extracted from the vibration experimental tests are 0.26 and 0.4 respectively under relaxed and contracted condition. The related mode shapes are extracted using the imaginary parts of the transfer functions and permit us to obtain a qualitative representation of the deformed mode shapes. Figure 5b gives a 3D representation of the imaginary part including the spatial dimension. This 3D representation give the deformed mode shapes of the volunteer at the two natural frequencies level. With this representation, the minima and maxima of the plotted surface can be easily related to their natural frequencies. It appears that, the first mode is associated with an extension motion and the second with a retraction motion, as shown in our previous study (Willinger*, et al.*, 2002) and illustrated in Figure 5a.

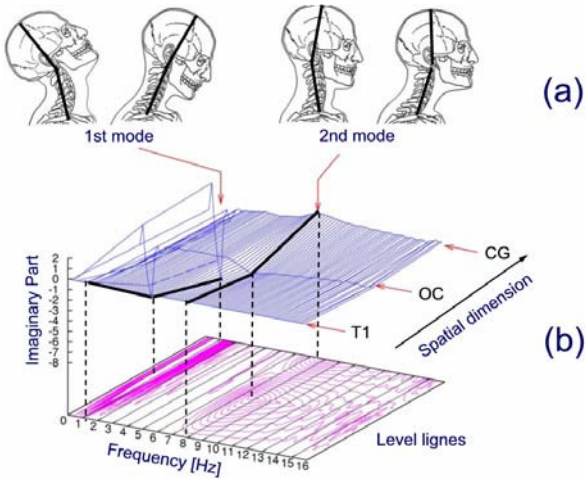
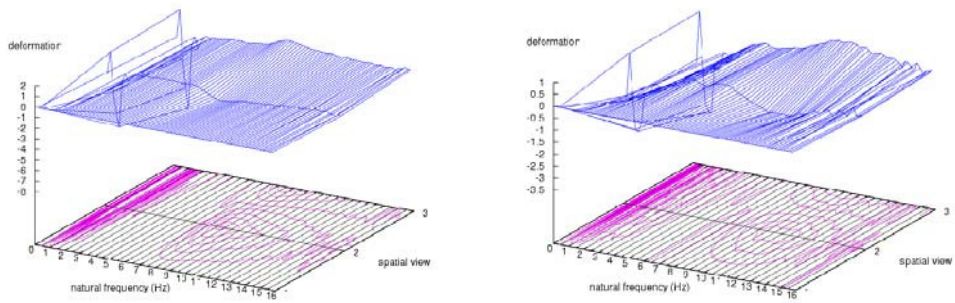


Figure 5 : Representation of the two deformed mode shapes and 3D representation of deformed mode shapes.



(a) (b)

Figure 6 : 3D representation of deformed mode shapes (a) in relaxed condition and (b) in contracted condition.

Figure 6 shows the 3D representations of the deformed mode shapes including spatial dimension under relaxed and contracted condition respectively. The main differences can be observed at the second natural frequency by an increasing of the transmittance amplitude between the relaxed and the contracted situation. This permits to conclude that mode shapes are similar for both experimental conditions.

INTER-INDIVIDUAL ANALYSIS

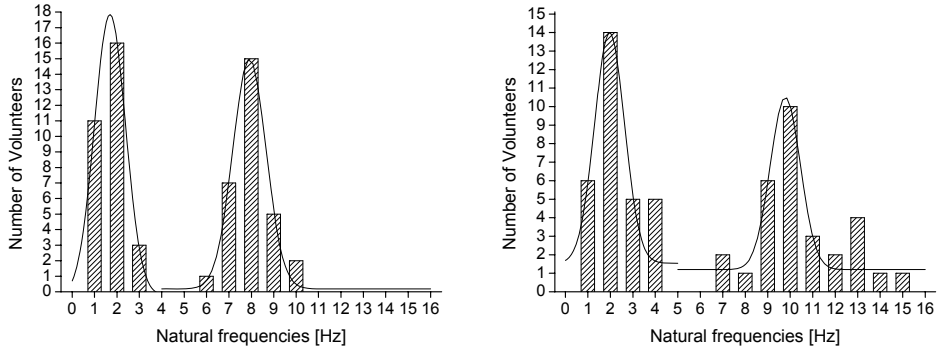
In order to analyze the inter-individual behavior, 30 volunteers were tested: 15 males and 15 females. The related anthropometric data are reported in Table 1. Results of these tests are synthesized in Table 2. For each volunteer both natural frequencies are extracted under relaxed and contracted condition and results are expressed in terms of histograms in Figure 7. statistical analysis show that the first natural frequency occurs at 1.68±0.69 Hz and the second at 7.91±0.74 Hz in relaxed condition (Figure 7a). Figure 7 b shows the histogram of the whole volunteers in contracted condition. We can clearly observe an increasing of both natural frequencies by about 15 % slipping from 1.68±0.69 Hz (relaxed condition) to 1.96±0.66 Hz (contracted condition) for the first natural frequency and of 21 % i.e. an increase from 7.91±0.74 Hz (relaxed condition) to 9.78±0.67 Hz (contracted condition) for the second mode. The damping ratios of the first mode could not be extracted because of a lack of energy at low frequencies as stipulated earlier. It was then impossible to compare the damping ratios between the two modes in this inter-individual study. On the other hand the previous study with an impulse modal analysis (Willinger *et al.*, 2002) showed that the damping ratios were virtually identical.

N°	Weight [kg]	Height [cm]	Neck circumf. [mm]	Neck length [mm]	Neck circumf. [mm]	Age [years]	Sex
1	75	183	370	130	580	34	M
2	51	156	310	115	540	34	F
3	51	151	315	120	560	25	F
4	79	186	360	135	565	23	M
5	78	171	400	120	610	33	M
6	90	170	400	130	590	25	M
7	52	170	320	155	550	36	F
8	52	158	320	115	550	40	F
9	53	155	320	120	560	30	F
10	104	182	415	130	560	52	M
11	74	172	390	130	575	47	M
12	57	161	345	125	556	36	F
13	63	158	320	125	560	30	F
14	82	190	390	135	610	36	M
15	97	169	440	125	590	45	M
16	71	176	360	115	580	24	M
17	57	161	345	125	556	33	F
18	75	178	365	125	580	35	M
19	85	171	350	120	575	38	F
20	65	160	360	115	540	35	F
21	63	168	350	130	550	30	M
22	67	177	320	130	540	27	F
23	57	158	325	130	515	34	F
24	55	168	360	130	550	53	M
25	79	165	385	115	580	38	F
26	89	175	400	135	580	60	M
27	76	178	400	140	590	58	M
28	52	156	300	130	550	36	F
29	64	165	330	120	560	39	F
30	81	176	400	120	585	55	M

Table 1 : Anthropometric data for the 30 volunteers.

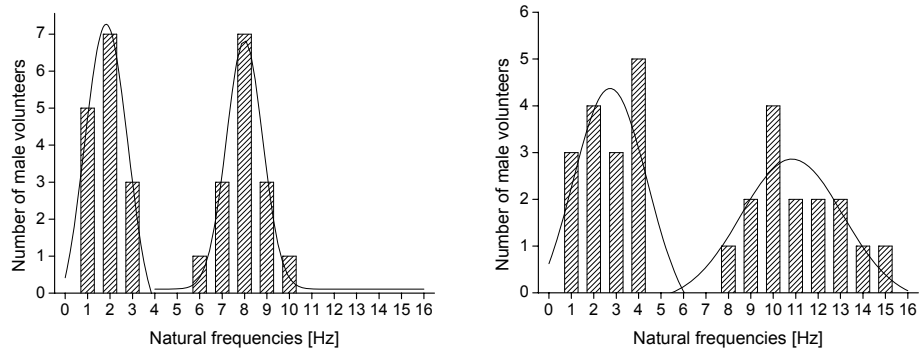
For the relaxed condition, the male and female dynamic behavior is almost the same, whereas for the contracted situation, the two groups distinguish itself clearly at the first mode, as illustrated in Table 1. Indeed, the relative variance between female and male groups for the first mode is of 17 % in relaxed situation against 34 % in contracted one. This can be explained by the fact that males have stronger neck muscles than females do. The male distribution is larger than the female one, so, there is more disparity in the male group.

Muscles activity has an effect on the neck dynamic response of the volunteer. For the male group, the first mode is affected by an increasing of the natural frequency of 40 % (1.83±0.96 Hz vs 2.7±1.66 Hz), whereas the second natural frequency increases by 30 % (8±0.8 Hz vs 10.8±2.27 Hz). The distribution of the male natural frequencies values in contracted condition are widespread, while the females' ones are closer, as illustrated in Figure 10. Moreover, for this group, muscle contraction affects both modes by an increasing of 20% (1.53±0.28 Hz vs 1.9±0.59 Hz for the first mode and 7.83±0.68 Hz vs 9.65±0.58 Hz for the second mode). Table 2 summarizes the results in terms of average natural frequencies with its standard deviation.



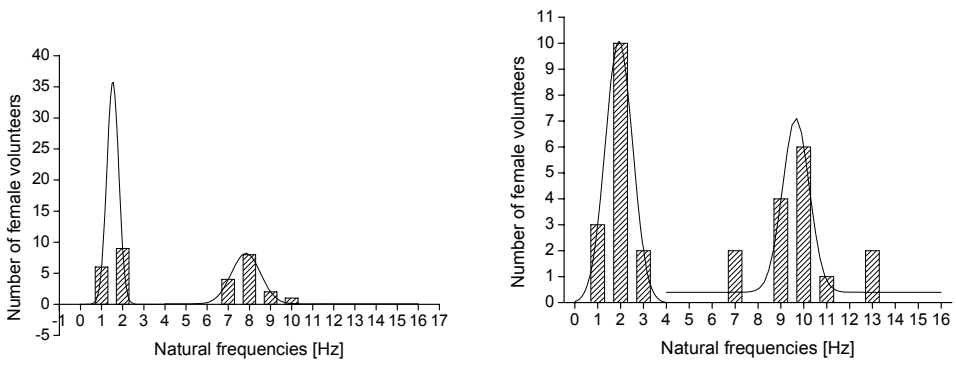
(a) **(b)**

Figure 7 : (a) Histogram of the whole volunteers in relaxed condition with its Gaussian curve; (b) histogram of the whole volunteers in contracted condition with its Gaussian curve.



(a) **(b)**

Figure 8 : (a) Histogram of the male volunteers in relaxed condition with its Gaussian curve; (b) histogram of the male volunteers in contracted condition with its Gaussian curve.



(a) **(b)**

Figure 9 : (a) Histogram of the female volunteers in relaxed condition with its Gaussian curve; (b) histogram of the female volunteers in contracted condition with its Gaussian curve.

Gender	Relaxed Condition		Contracted Condition	
	f_1	f_2	f_1	f_2
Female	1.53 ± 0.28	7.83 ± 0.68	1.9 ± 0.59	9.65 ± 0.58
Male	1.83 ± 0.96	8 ± 0.8	2.7 ± 1.66	10.8 ± 2.27
Human	1.68 ± 0.69	7.91 ± 0.74	1.96 ± 0.66	9.78 ± 0.67

Table 2 : Results of female and male groups and the whole volunteers in terms of natural frequency with its standard deviation under relaxed and contracted condition.

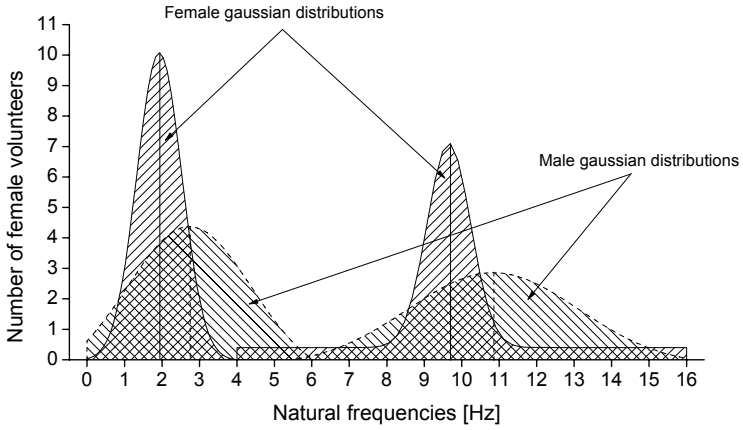


Figure 10 : Superimposition of the Gaussian distributions of the two natural frequencies recorded for the female and male group under relaxed conditions.

DISCUSSION

In the present study two configurations have been conduced on 30 male and female volunteers in relaxed and contracted condition. In case of relaxed situation, the results of male and females volunteers are very similar (f_1=1.83±0.96 Hz and f_2=8±0.8 Hz for male group and f_1=1.53±0.28 Hz and f_2=7.83±0.68 Hz for female group). Under contracted condition, male volunteers present higher values of natural frequencies of 17 % for the first mode and of 2 % for the second mode in comparison with femals. The major difference is clearly about the first natural frequency. This can be explained by the fact that neck muscles are stronger for male than for female and this muscles affect principally the flexion-extension motion, which are motions of everyday life. Indeed, the relative variance between male and female volunteers at the first natural frequency is of 34% (1.9±0.59 Hz for female group vs 2.7±1.66 Hz for male group), while the one for the second mode is only of 11 % (9.65±0.58 Hz for female vs 10.8±2.27 Hz for male). These results are in good accordance with the previous impulse modal analysis results. It should be noted however that this study is restricted to human males and females groups with no separates groups as size and stoutness. The linearity assumption also excludes severe rear impact analysis with major neck extension and investigation of tolerance limits.

Extracted deformation mode shapes have often been observed qualitatively in the literature, when volunteers were investigated in the temporal domain under rear impact condition. It has been reported that the retraction appears “a few milliseconds” before neck extension (Boström *et al.*, 1997, Deng and Goldsmith, 1987, Kleinberger, 1993, Ono and Kaneoka, 1997, Walz and Muser, 1995). These findings are consistent with the present modal analysis, although to our knowledge no study has clearly defined the conditions under which this retraction mode does or does not appear. This study leads us to the conclusion that the retraction mode is only excited if energy is introduced into the system above 8 Hz. This applies only if the impact duration is sufficiently short or if high loading ramps exist within the loading function. This is in complete agreement with Nightingale’s findings following investigation of the neck under vertical loading with a multi-body model restricted to the temporal domain Nightingale *et al.*, 2000. The main result was that faster loading rates were associated with high order buckling modes. In addition, these authors reported that injury mechanisms may be substantially changed by loading rates as inertial effects may influence whether or not the cervical spine fails in compression mode, or in bending mode in their case. In terms of modal analysis this statement simply becomes “if a

natural frequency and its mode shape is excited, the related injury mechanism is potentially present". The present study therefore provides new insight into injury mechanisms affecting the upper and lower cervical column following rear end impact. In this study, we did not study the rebound after the rear end impact which appears 200 to 300 ms after the beginning of the impact, given that our model was not validated up to this duration but only up to 150 ms. Moreover there is not yet clear evidence if the rebound is a critical event in low speed rear impact accidents.

The main methodological question is the assumption of linearity arising from the definition of transfer function and also the low impact energy involved in the experimental impact. It is important therefore to remember that the methodology is well designed to describe neck behavior in low energy impact or before non-linearity due to saturation (hyperextension of ligaments, bone contact, muscle activity) occurs. Indeed, in our tests, the head and neck rotations do not exceed ten degrees. These allow us to characterize the head-neck system only in the linear domain for low speed rear impact. It is questionable whether the neck really exhibits non-linear behavior under low speed rear impact. Bogduk and Yoganandan, 2001, Matsushita *et al.*, 1994, McConnell *et al.*, 1993, Yoganandan *et al.*, 1995 have stated that the head does not rotate beyond physiological limits. In the real world, accident injuries are also often seen even when the head rest is present, under low energy, showing that it can occur early after impact before extension of the head takes place and probably before non-linear behavior appears. This illustrates that injuries may occur under low displacement conditions.

On the other hand, linear behavior has been confirmed in our experiments by the coherence function which remains between 0.9 and 1. The results illustrate the importance of mass distribution and damped elastic properties at the beginning of the motion. Resonance frequencies and mode shapes provide initialization for the dynamic deformation that may eventually continue until non-linearity appears in case of energetic impact. The definition of mode shape hinders our understanding of the mechanisms of injury even if movements are restricted to small deformations. In our case the extracted modal characteristics of the neck represent "Eigen" mechanical data for the head-neck system in the sagittal plane whatever the loading conditions. New validation parameters are therefore determined for a model, restricted to low energy impacts, remaining in the sagittal plane. On the other hand the response of the temporal model will clearly depend on loading conditions such as location of head impact, impact force time history or inertial head loading due to first thoracic vertebra acceleration in the antero-posterior direction

CONCLUSION

In the present study, 15 human females and 15 human males head neck systems have been subjected to a frequency analysis under thorax vibration. This work aims at extracting natural frequencies of the head-neck system associated with deformed mode shapes on a larger population. It completes the previous study and supports the previous results found by impacting the forehead of the volunteer. The 30 human volunteers seated in a rigid seat were submitted to a horizontal vibration applied to a carriage. Their thorax is rigidly maintained in order to prevent any relative movement between T1 and the seat backrest. Neck characterization is performed in the frequency domain by a vibration loading of the volunteer thorax. Inter-individual differences, gender influence and muscle effect are also investigated. The mean modal characteristics of the human head-neck system extracted from the 30 experiments and the related standard deviation are defined by two natural frequencies located at f_1=1.68±0.69 Hz and f_2=7.91±0.74 Hz. For all subjects the first vibration mode was a flexion-extension motion and the deformed mode shape related to the second natural frequency was a head translation or retraction motion. These results are in good accordance with the previous results that has been conduced by Willinger, *et al.*, 2002, who impact frontally the forehead of five volunteers in order to practice a modal analysis of the head-neck system *in vivo*. When male and female were distinguished, only small differences appeared in relaxed condition, i.e. for the male, mean value and standard deviation of the two natural frequencies of f_1=1.83±0.96 Hz and f_2=8±0.8 Hz against f_1=1.53±0.28 Hz and f_2=7.83±0.68 Hz for the female group. The contacted condition shows a major increasing of the first natural frequency value. The two natural frequencies for both male and female groups are respectively f_1=2.7±1.66 Hz and f_2=10.8±2.27 Hz against f_1=1.9±0.59 Hz and f_2=9.65±0.58 Hz.

Moreover, it was shown in this study that inter-individual differences and gender influence is low, illustrating that each head-neck system has his own set of mechanical properties which lead to a

similar dynamical behavior. The new data recorded on human being in vivo constitute new validation parameters for dummy neck evaluation and mathematical models validation. In further research the reported modal behavior will conduce to new possibilities in neck protective systems evaluation and optimization in the frequency domain.

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