COMPARISON OF MANUAL WHOLE-BODY AND PASSIVE AND ACTIVE HEAD-ON-BODY ROTATIONAL TESTING WITH CONVENTIONAL ROTARY CHAIR TESTING

Jason M. Hanson and Joel A. Goebel

Department of Otolaryngology—Head and Neck Surgery, Washington University
School of Medicine, Saint Louis, Missouri

Reprint address: Joel A. Goebel, MD, FACS, Department of Otolaryngology—Head and Neck Surgery, Washington University School of Medicine, 517 S. Euclid Avenue, Box 8115, Saint Louis, MO 63110.

Tel: (314) 362-7561; Fax: (314) 367-7346; E-mail: joel@vertige.wustl.edu

Abstract—New methods of rotational testing of the vestibulo-ocular reflex (VOR) using manually generated or patient-generated sinusoidal head movements have recently been advocated for clinical use in circumstances where conventional rotary chair testing methods are not feasible. However, studies seeking to provide evidence for the validity of these methods by comparing their results with an accepted “gold-standard” have been conspicuously absent in the literature. In this study, we compared results for VOR gain, phase, and asymmetry obtained using whole-body and head-on-body rotational stimuli with those obtained by conventional rotary chair testing in 35 subjects with either unilateral or bilateral vestibular deficits over the range of frequencies from 0.025 Hz to 1 Hz. Our results provide evidence for the validity of manual whole-body and active and passive head-on-body rotational testing methods by demonstrating excellent agreement between each of these and conventional rotational chair testing for VOR gain, phase, and asymmetry within the frequency range studied. Small differences at specific-paradigm datapoints are likely secondary to subtle limitations of our experimental design. With further refinement, we expect the new methods will be useful adjuncts for evaluating patients with vestibular complaints in selected clinical situations. © 1998 Elsevier Science Inc.

Introduction

Rotational testing of the vestibulo-ocular reflex (VOR) is an important part of the evaluation of patients with complaints of dizziness, vertigo, and dysequilibrium (1). The currently accepted “gold-standard” method of testing employs a servo-controlled rotary chair apparatus for applying a sinusoidal whole-body rotational stimulus to determine the gain, phase, and symmetry of the VOR. While this standard method is designed to obtain highly reproducible results under optimal conditions, its application and availability for widespread clinical use are limited by the size and expense of the equipment required and the manner in which testing must be performed.

Recently, new methods of rotational testing have been described in the literature that require minimal equipment, are lower in cost, and allow greater flexibility of testing by permitting manually generated or patient-generated rotational stimuli in a variety of clinical settings (2–8). We describe in this article the application of these methods in our laboratory using a rotational testing system that allows assessment of the VOR using three stimulus types: manual whole-body-rotation (similar to conventional rotary chair testing), passive (examiner-generated) head-on-body rotation, and active (patient-generated) head-on-body rotation. Previous work in our laboratory has documented our use of these methods.
in selected individuals at 0.5 and 1 Hz (9-11). Refinements in our methods now permit evaluation of the VOR using rotational frequencies from 0.025 Hz to 1 Hz for whole-body rotation and from 0.25 to 1 Hz for head-on-body rotation.

Evidence has been presented in the literature for the reliability (i.e., reproducibility of results) of systems employing new rotational testing methods (12). However, validation of these new methods by comparison of their results to an accepted “gold-standard” is critical before their acceptance for widespread clinical use. Such comparisons are conspicuously absent in published reports.

To provide evidence for the validity of these methods, we compared results for VOR gain, phase, and asymmetry obtained using whole-body and head-on-body rotational stimuli with those obtained on the same day by conventional rotary chair testing in 35 subjects referred to our clinical laboratory for evaluation of vestibular deficits.

Methods

Subjects

Thirty-five patients (mean age 57.8 years, range 32 to 77 years) with complaints of dizziness, vertigo, or dysequilibrium and documented caloric deficits were enrolled in this study. Subjects were categorized as having either unilateral or bilateral vestibular deficits based on results of caloric testing. A unilateral vestibular deficit was defined as greater than 30% asymmetry for binaural bithermal closed-loop irrigations according to the vestibular paresis formula:

\[
\% \text{ paresis} = \frac{(\text{Right}_{\text{warm}} + \text{Right}_{\text{cool}}) - (\text{Left}_{\text{warm}} + \text{Left}_{\text{cool}})}{\text{Right}_{\text{warm}} + \text{Right}_{\text{cool}} + \text{Left}_{\text{warm}} + \text{Left}_{\text{cool}}} \times 100\%.
\]

A bilateral vestibular deficit was defined as the sum of the four irrigations less than 20%/s and less than 30% asymmetry according to the formula above. Of the patients, 21 were men and 14 were women. Twenty-one patients were classified as having unilateral vestibular deficits: 14 had bilateral deficits. Presumed causes of vestibulopathy were idiopathic (16 patients), postsurgical unilateral deficit (8 patients), acoustic tu-
signal from which fast components were identified and removed. Additional manual editing of saccades was performed as necessary. Cycles with poor signal quality were manually rejected, and the remaining cycles were overlaid to create a composite signal. The gain, phase, and asymmetry (DC bias/response amplitude) of the fundamental of this composite (relative to velocity) were determined by Fourier analysis. Correction for head slippage within the headholder at the highest frequencies was made according to an algorithm developed in our laboratory (see Discussion).

New Methods of Rotational Vestibular Testing

Immediately after rotary chair testing, subjects were taken into an adjoining room where the experimental rotational vestibular testing system was employed. Each subject was seated in a standard examination chair which could be freely rotated. Eye movements were recorded using the same EOG electrode placement. Calibration of the EOG signal was performed initially and at every other trial using ±20 degree saccades with the head held stationary at the center position. Head movements were measured by an angular rate sensor (Watson Ind., Madison, WI) mounted on an adjustable plastic headband fit snugly around each subject's head. Care was taken to ensure that the angular rate sensor moved in the yaw plane at all times and that translational movements did not occur.

Subjects were tested using manual whole-body rotation, passive head-on-body rotation (for which the examiner smoothly rotated the subject's head), and active head-on-body rotation (for which the subjects smoothly rotated their heads). An audible cue generated by the testing system was used by the examiner for passive movement or by the subject for active movement in timing head rotations. For whole-body rotation trials, the same six frequencies as for rotational chair testing were employed (0.025, 0.05, 0.1, 0.25, 0.5, and 1 Hz), with the same approximate peak velocities (25°/s for 0.025 Hz, 50°/s for remaining frequencies). Active and passive head-on-body rotation was performed at 0.25, 0.5, and 1 Hz. On-line monitoring of head movement was used to evaluate the timing and smoothness of the generated head movement. Testing was performed in the presence of optokinetic stripes for the visual–vestibular interaction (Visual–VOR) paradigm, and in a dimly lit room with the subject's field of view covered with an opaque visor for the VOR–darkness paradigm. Subjects received continuous reinforcement throughout all trials to ensure alertness.

Head and eye movement data were collected and analyzed using a testing system being developed in this laboratory in cooperation with Nicolet Inc. (Madison, WI) consisting of an 80486-class desktop computer with 4 megabytes of RAM. For each trial, the EOG signal was low-pass filtered (DC-24 Hz) and sampled at 200 Hz. The eye position record was digitally differentiated to obtain an eye velocity signal, from which fast components were removed by software. Additional manual editing of saccades was not possible. Individual cycles were identified utilizing zero-crossings of the eye velocity record. Since actual cycle frequency varied around the target frequency, the gain, phase, and asymmetry of the response for each cycle were determined by Fourier analysis. Cycles with poor signal quality, defined as >40% total harmonic distortion of the stimulus or response, were rejected. The mean values for gain, phase and asymmetry of the remaining cycles were then calculated.

Statistical Methods

We used a repeated-measures analysis of variance and Dunnett's T-test to compare each of the manual system's testing methods to the standard of the conventional rotational chair when segregated by test paradigm (VOR–darkness, Visual–VOR) and rotational frequency. We chose an alpha level of 0.05 as significant.

Results

Raw Data

Representative data tracings for one subject are shown in Figure 1. Note that the conven-
Figure 1. Comparison of composite conventional rotary chair tracing (top left) with a representative cycle from the manual whole-body rotation test (bottom left), and the passive (top right) and active (bottom right) head-on-body rotation tests at 0.5 Hz with a peak velocity of 5°C/s for the VOR-darkness paradigm. Head velocity (H), horizontal eye velocity (E), and Fitted Curve are as labeled. Note that the conventional system computes VOR parameters from a single composite cycle, whereas the experimental system computes these parameters for each cycle then averages the results. Amplitude and timing measurements for examiner- and patient-generated movements are shown with each cycle. VOR parameters for the experimental system are referenced against the actual head movement.
tional system computed VOR parameters from a single composite cycle, whereas the experimental system computed these parameters for each cycle and then averaged the results.

The experimental testing system allowed for stimulus frequencies of up to 2 Hz. However, early in our series of patients, we noted that limitations of the head restraint device on the conventional rotary chair did not allow for rigid fixation during testing at 2 Hz, and the resultant data showed a high degree of artifact. We therefore chose to limit our analysis to frequencies of 1 Hz and below, where meaningful comparisons could be made. Additionally, most patients were not able to generate a smooth sinusoidal head-on-body rotation at 0.25 Hz, even with proprioceptive cues. Therefore, this data point (active head-on-body rotation, 0.25 Hz) was also excluded from analysis.

**VOR Gain**

Results for gain calculations for each testing method are shown in Figure 2 for both the VOR-darkness and Visual-VOR paradigms. Results for the new rotation test methods (open symbols) are compared to those of the conventional rotary chair test (filled symbols). Only a few small differences existed between test methods for either test paradigm. For the VOR-darkness paradigm, gain for manual whole-body rotation was statistically less than rotary chair testing at 0.05 and 0.1 Hz. The difference between means was approximately 0.05 gain units. For the Visual-VOR paradigm, gain for manual whole-body rotation was statistically less than rotary chair testing at 0.025 and 0.05 Hz. The difference between means ranged from 0.10 to 0.15 gain units. Passive head-on-body rotation was statistically greater than rotary chair testing at 0.25 Hz by 0.09 gain units for the Visual-VOR paradigm only.

**VOR Phase**

Results for phase calculations are shown in Figure 3. For the VOR-darkness paradigm, rotary chair phase was statistically less than all three of the new rotational tests only at 1 Hz. For the Visual-VOR paradigm, small standard deviations led to significant differences 0.25 and 0.5 Hz. Neither of these differences, however, exceeded 5 degrees. At 1 Hz there was a larger statistically significant difference between the rotational chair test and all three of the new rotational methods. For both paradigms, the experimental method's phase results converged to the expected result of zero phase as frequency increased, whereas rotary chair phase results showed an increasing phase lag.

**VOR Asymmetry**

Results for asymmetry calculations are shown in Figure 4. There were no significant differences between the experimental methods and rotary chair testing for either paradigm at any frequency.

**Discussion**

This study provides strong evidence for the validity of the new methods of rotational testing of the vestibulo-ocular reflex (VOR) described in this report over the frequency range studied. We feel that many of the small yet statistically significant differences between test methods are attributable to specific limitations of our experimental design.

Gain differences between the test methods were found only at low frequencies of rotation for both the Visual-VOR and the VOR-darkness paradigms. Examination of the raw data records showed that imperfect saccade rejection by the experimental system's analysis software led to a systematic underestimation of VOR gain for frequencies of rotation wherein a large number of re-centering saccades occurred, namely 0.025 to 0.1 Hz. Refinements in our analysis algorithm will likely reduce or eliminate these differences. In addition, the noticeably less compelling optokinetic stimulus (stripes projected onto the walls of a standard examining room) used with the experimental system likely led to an additional gain decrement for the Visual-VOR paradigm (0.10 to 0.15 gain units for Visual-VOR versus 0.05 for VOR-darkness).
Small yet statistically significant phase differences across the range of frequencies tested may reflect idiosyncratic differences between analysis algorithms. The statistical power of our analysis is demonstrated quite strikingly at 0.5 Hz under the Visual–VOR paradigm, where differences between means of 2 and 3 degrees are determined to be statistically significant owing to our large n and the particular distribution of phase values at this frequency–paradigm combination. However, it is unlikely that such small differences would be considered clinically relevant.

Figure 2. Gain comparisons for the VOR–darkness (A) and Visual–VOR (B) conditions. Frequency is indicated along the X axis. Gain, in dimensionless units, is on the Y axis. Mean values across subjects are indicated by each data point for the conventional rotational chair (black circles), manual whole-body rotation (squares), passive head-on-body rotation (triangles), and active head-on-body rotation (diamonds). Significant differences (P ≤ 0.05) between the experimental methods and rotational chair testing are indicated by an asterisk. Lesser gain values for the experimental system at lower frequencies is felt in part to be secondary to imperfect saccade editing and the less compelling optokinetic stimulus presented. See text for further discussion.
The larger phase differences that occurred at 1 Hz for both the Visual-VOR and the VOR-darkness paradigms are felt to reflect systematic error arising from head slippage within the rotary chair's head-holder. We have documented similar systematic phase differences in normal subjects undergoing conventional rotational chair testing that led to overestimation of VOR phase (13). It may be argued that vestibularly deficient subjects will have less head stabilization ability (and therefore a measurably greater degree of head slippage) when rotated at these frequencies.
frequencies. The use of a rigidly mounted head rotation sensor during higher frequency rotary chair testing could quantify this error.

There were no statistically significant differences for measurements of asymmetry. We did note divergence of asymmetry values for whole-body rotation and conventional rotary chair for the Visual–VOR paradigm (optokinetic surround) which was not present in the VOR-darkness paradigm. A difference in the specific visual stimuli for the two methods (as noted in the gain data) may be responsible for this observation. For frequencies of rotation of 0.25 Hz and greater wherein a subject could fixate (or imagine fix-

![Asymmetry comparisons for the VOR-darkness (above) and Visual–VOR (below) conditions. Percent asymmetry is indicated on the Y axis. None of these differences are significant ($P > 0.05$).](image-url)
demonstrated no significant differences between the techniques are clinically significant and, if so, whether they can be reduced or avoided by these refinements. Important considerations in this regard are cervical input, volition, elimination of visual targets, irregularities in the rotational stimulus, and the technical challenges of analyzing two irregular waveforms.

The cervico-ocular reflex in humans is thought to play a limited role in gaze stabilization under healthy conditions (14). Although no systematic study has shown enhancement of VOR gain during head-on-body versus whole-body rotation in vestibulopathic patients, isolated case reports have described individuals with presumably strong cervical contributions to VOR gain during head rotation (15). Prior studies in our laboratory failed to show a consistent difference between head-on-body and whole body VOR gain in normal subjects (9) or in patients with reduced vestibular responsiveness on caloric testing (11). While one statistically significant enhancement with head-on-body rotation was noted (gain at 0.25 Hz, Visual-VOR), this study demonstrates no consistent overall VOR enhancement with neck rotation. Further evaluation of a larger number of subjects is necessary, however, before it can be definitely determined that cervical inputs do or do not enhance the VOR during head-on-body testing in vestibularly deficient patients.

Volition plays a theoretical role between active and passive head-on-body rotations because in one instance the movement is self-generated and in the other it is not. Furman and colleagues noted no appreciable differences between active and passive head rotations from 1 to 5 Hz (16). Our previous two studies also demonstrated no significant differences between the two conditions, although a trend towards higher VOR gain was noted on the active condition (9, 11). It is possible that certain vestibulopathic patients may utilize preprogrammed eye movements during volitional head turns, but, as a whole, this trend has not been convincingly demonstrated.

The influence of visual targets on VOR gain and phase is most pronounced at frequencies below 1.0 Hz. In this frequency range, viewing a stationary target during head movement enhances the VOR by mechanisms thought to be related to smooth pursuit and optokinetic stimulation (17). Conversely, viewing a target moving in tandem with the subject suppresses VOR-generated movements in order to maintain fixation. Above 1.0 Hz, however, pursuit and optokinetic gain decrease rapidly and the vestibular reflexes dominate gaze stabilization during head movement. Since our study focused on studying the VOR below 1.0 Hz, we chose to eliminate visual fixation with a shield before the eyes to simulate testing in darkness. Our previous work demonstrated no significant difference in VOR gain in the rotary chair at 0.5 and 1.0 Hz in normal subjects tested in the dark versus in the light with the shield before their eyes (9). We feel confident, therefore, that this technique adequately eliminates visual fixation and simulates dark-field VOR conditions.

The technical issues of waveform analysis and artifact rejection during this type of rotational testing represent perhaps the greatest challenge. With rotary chair testing, a smooth and accurate sinusoidal stimulus is delivered to the patient from the computer algorithm coupled to a powerful servo-controlled motor. With manual whole-body or head-on-body testing, however, both the stimulus and the eye movement velocity tracings are not regular, and the computer algorithm must carefully estimate, cycle by cycle, the actual frequency and peak velocity of both signals. Identifying and rejecting artifacts accurately for accepting or rejecting individual cycles is difficult yet critical. We believe this study demonstrates the capability of manual testing to overcome these technical obstacles, at least within the frequencies and peak velocities examined. At frequencies above 2 Hz and velocities greater than 100°/s, compensatory nonvestibular eye movements may be confused with VOR signals, complicating the analysis.

The next phase in this investigation of new methods of VOR testing must include subjects
with markedly decreased vestibular function at all frequencies of rotation to better determine whether cervical input or volition covertly enhance VOR gain and phase. Refinements in stimulus delivery and eye movement recording with video techniques will further improve this type of vestibular testing. The ultimate goal of this project is a fully portable, broad-frequency (0.025 Hz to 6 Hz) system that will allow both office and bedside testing, diagnosis, and monitoring of vestibular function.

Acknowledgments — We wish to acknowledge Keith Dunnigan, PhD, for assistance with the statistical analysis of the presented data and Laum Langhofer, BS, for software development for the manual rotational testing system. Supported in part by NIH Grant T-32 DC00022.

REFERENCES